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A biomechanical study of equine locomotion /

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A BIOMECHANICAL STUDY OF EQUINE LOCOMOTION

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1. Abstract

The conditions produced by the impact of the horse's hoof with the ground were studied. Accelerometers attached to the hoof were used to measure the shock waves generated by impact under varying conditions. Analog signals from the accelerometers were recorded by a small tape unit carried by the rider. After digital conversion, computer analysis of the impact shock waves was performed in time and frequency domains. Averaging techniques were used to include from fifty to one hundred contiguous strides. Peak impact acceleration was found to vary with gait, speed, and horseshoe pad material. Power density of impact acceleration was found to vary with gait, horseshoe pad material, and ground surface. Using multiple accelerometers, gait patterns were recorded from different points and intraskeletal shock attenuation was measured.

2. INTRODUCTION

The equine locomotion system is an impressive product of functional design. The legs have been optimized in terms of skeletal weight and strength, angular velocity, stride frequency, and energy return. The resulting combination of speed and stamina is unmatched by any animal of this size.

Studies have been presented in the literature concerning the internal and external forces which act on the equine lower limb during the support phase. Using a variety of *in vivo* and *in vitro* methods, researchers have been able to determine the character of these forces with regards to magnitude, direction, duration, and rate of application.

Photography is the oldest of the gait analysis methods available, going back to the work of Muybridge (1887). His study used a sequence of still pictures and was contracted to settle a bet concerning the existence of a flight phase at the gallop. This was the first time that a method of gait analysis had sufficient resolution at high speed to prove this point.

Current photographic methods use high speed cameras to correlate other quantitative data. Kingsbury et al. (1978), Merkens et al. (1985), Schamhardt and Merkens (1987), and others have used the photographic record to define the features of temporal force plate data. Force plate traces are characterized by certain features which are repeated with every stride. Photography is used to relate some of these force-time plot features to the corresponding physical event.

Bartel et al. (1978) combined force plate measurements and photography for calculating the internal forces in the lower leg. After the dimensions of the force members had been obtained through X-rays and the applied force measured with a force plate, photographic analysis was used to determine the orientation of the leg at a number of intervals throughout the support phase. The union of this data provided enough information to allow computation of the internal forces in the digit.

A good deal of force plate research has focused on finding a number of indices to be used in gait comparison. Since diagnosis of equine lameness is often reduced to a trial and error process, the existence of a reliable diagnostic tool would be invaluable. Merkens et al. (1985), and Schamhardt and Merkens (1987) perform such work with encouraging results. One of the obvious shotrcomings of the force plate approach to this problem is the lack of continuous data. The horse is led over the force plate and the impact and support of one or two legs is recorded. Successive trials are obtained by leading the horse back and forth over the stationary plate. A better approach might be to record a large number of successive strides, but this suggests an extremely large force plate.

Many practicable results can be produced by on-board instrumentation. Not limited by the size of the force plate, researchers using the on-board approach may obtain data from contiguous support phases. In addition to obtaining continuous data, the on-board approach provides researchers a second important advantage. This is that once the horse is instrumented, he may be worked over any surface and at any gait. This means that the study can investigate the conditions surrounding the natural performance of the animal. This has the effect of taking the research out of the lab, and getting it into the real world.

Frederick and Henderson (1970) incorporated pressure transducers in horseshoes to measure ground reaction force. They presented data from successive support phases from a test horse at three gaits and on different surfaces. Their results are prominent in the literature and are often used as a benchmark for comparisons of subsequent results. If the methods of Frederick and Henderson (1970) lacked anything, it was equipment. Although they had modern equipment for their time, their data output was in the form of paper strip charts it required a small motorized cart to lug the apparatus around.

Nunamaker et al. (1986) took advantage of size and weight reductions of measuring and recording instruments to gather data for fatigue fracture tests. On-board recording of the *in vivo* strain of the third metacarpal of Thoroughbreds at training and racing speeds shows that the third metacarpal at impact is in bending, as well as compression, and that the direction of the principal axes is not constant. This work holds tremendous promise of applicability since roughly 70% of young Thoroughbreds in training suffer from fatigue fractures of the third metacarpal bone.

The investigation of impact has not been studied as thoroughly as the rest of the support phase of the stride. Many researchers routinely cover the force plates with a rubber mat to eliminate slipping and to reduce the amplitude of the impact spike. Schamhardt and Merkens (1987) used linear extrapolation to eliminate the transient impact from their temporal force data. The resolution of most force plates is designed to be optimum during mid-stance. Such instrumentation is inadequate to study impact.

While thorough impact studies of equine locomotion are lacking in the literature, ample evidence is given to show that the strain, strain rate, and shock waves generated by impact have substantial effect on damage to the horse. Behiri and Bonfield (1984) studied the effects on the fracture mechanics parameters of bone and found there to be a transition point between slow stable bone crack propogation and catastrophic failure. Their conclusions agree with the findings of others such as McElhaney and Beyers (1965), Lanyon and Smith (1970), and Lanyon (1971). In these cases the dependence of the amount of bone damage with the applied strain rate was quite dramatic.

In addition to the application of strain to the bones of the lower leg, the impact phase is characterised by the generation of shock waves upon heel strike. As the horse is viewed as a product of evolutionary design, the damping and shock absorbing mechanisms of the legs can be seen to be primary defenses against skeletal damage.

Rooney (1979), Kingsbury et al. (1978), and Pratt and O'Connor (1976) discuss the fact that the first 20% of *in vivo* force plate readings are influenced by the heel strike. Pratt and O'Connor (1976) model the heel strike as an underdamped impulsive response superposed onto the vertical support force. Kingsbury et al. (1978) note the absence of such a feature in force traces from dead legs and suggest that the difference is due to the shock absorbing capacity of live tissue and the shock absorbing functions of the musculature of the leg.

All three of these studies agree on the fact that one of the most important functions of the leg is control of impact energy by dissipation, attenuation, and conversion. Pratt and O'Connor (1976) also note that in tired and lame animals the impact portion of the force curve becomes accentuated. They theorize that loss of muscular control over impact may be related to injury. Similarly, Rooney (1979) concludes that fatigue in the horse which leads to damage of the leg is partly a result of muscular fatigue leading to a loss of vibrational damping.

There are several design features of the equine locomotion system which serve to redirect, dissipate, and convert the energy of impact. Pratt and O'Connor (1976) discuss conversion of kinetic energy to potential energy by closing of the shoulder and stifle joints, and by sinking of the fetlock. Hildebrand (1987) presents a thorough report on inertia and energy as they relate to the biological design criteria and solutions of the locomotion system of the horse. Evolutional selection of the most feasible methods of impact shock control has resulted in a remarkable system.

Since impact shock has influenced the natural design of the horse, and the forces at impact have been proven to be controlling factors in the damage sustained by the system, it is natural that this topic be chosen for study. What has been previously lacking was a combination of techniques which would allow measurement of impact shock, reliable mounting of the measuring devise, and on-board recording.

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It is the purpose of this research to develop methods for investigation of the conditions which are produced by impact of the hoof with the ground. The experimental techniques used are an extension of the methodology used successfully by Loy (1987) and Loy and Voloshin (1987) in work performed at the Biomechanics Division of the Department of Mechanical Engineering and Mechanics, Lehigh University. These studies used skin mounted accelerometers with human subjects to characterize impact under varying conditions and to investigate the effectiveness of means and materials for shock attenuation.

While the investigation of equine locomotion differs somewhat from experiments in human locomotion, the data collection systems in each case posses the following qualities which are essential to the study of biomechanical impact:

- provide reliable and accurate data
- have sufficient resolution at high speed
- be sensitive to differences in impacts
- be easily attached to the test subject
- feel natural to the subject so as not to change the gait patterns in any way

The accelerometric system was chosen because it satisfies all of these requirements at reasonable cost.

There are three main sections in this report, each containing two or more chapters. Chapters 3 and 4 describe the nature of equine locomotion and the forces to which the leg is subject. Knowledge of the dynamic system and its operating conditions is crucial to experiment design and interpretation of results. Chapters 5-10 discuss measurement theory, experiment design, and data collection. The ultimate goal of this study was the development of this methodology for investigation of impact generated shock waves in horses. The final section, chapters 11-14, presents results gained through these investigations.

3. The Natural Gaits

The horse has three natural gaits, all of which are investigated here. It has been suggested by Hildebrand (1965) that the gaits are not distinct but rather are part of a continuum of possible forms of locomotion. However, convenience and standard practice dictate that they be considered as separate.

Diagramatic representations of gait will be helpful to the following discussion and will be presented next. These footfall diagrams have circles indicating the limbs which are in the support phase. The view is that of an observer looking down on the horse as the horse moves from the left to the right.

The walk is an even four-beat gait. The limbs contact the ground in precise intervals (Fig. 1). The impacts occur at the beginning of phases 2,4,6 and 8. Each of the limbs spend more than half of the limb cycle in support. There is no time during the walking cycle where there are less than two legs in support.

```
  o            o   o       o    o      o    o
1 --------> 2 --------> 3 --------> 4 -------->
       o            o                  o

          o           o                  o
  5 --------> 6 --------> 7 --------> 8 -------->
    o           o    o       o    o      o  o
```

Fig. 1. Footfall diagram of the walk.

The trot is an even two-beat gait in which the limbs move in diagonal pairs (Fig. 2). Note from the footfall diagram that there is a suspension phase where none of the limbs are in contact with the ground and where the horse is in ballistic flight.

```
  o                              o
1 -------->  2 -------->  3 -------->  4 -------->
        o                    o
```

Fig. 2. Footfall diagram of the trot.

```
  o            o                              o
1 -------->  2 -------->  3 -------->  4 -------->
               o            o            o

        o            o
5 -------->  6 -------->  7 -------->  8 -------->
                     o            o
```

Fig. 3. Footfall diagram of the right lead transverse gallop.

Like the trot, the gallop has a suspension phase but it is not an even gait. Properly named the transverse gallop, there exist both right lead and left lead gallops depending on the order in which the limbs are placed on the ground. The legs of a horse galloping on the right lead (Fig. 3) will land in the order of: left hind, right hind, left fore, right fore. On the left lead the order will be: right hind, left hind, right fore, left fore.

The canter is a restrained gallop. It is the three-beat gait which occurs when the lead hind leg and the nonlead fore leg form a diagonal pair. In either case the lead leg is the last of a contralateral limb pair to leave the ground. It is this lateral support by the lead foreleg which makes the horse readliy lead with the innermost foreleg while turning. The support forces and impact forces vary between the lead and non-lead foreleg. Impact forces will be seen to vary in terms of both peak acceleration and power density of acceleration.

4. Forces on the the Lower Limb

The support phase of the stride can be divided into three parts. The first phase, impact, is the time from initial contact with the ground to the time when the hoof is firmly seated on the surface. The deceleration phase of the limb cycle is from the end of impact to mid-stance. Propulsion, the third phase is from mid-stance to toe-off. These divisions are not always absolute. On slippery surfaces or on a surface with a low shear strength, it is hard to determine the end of the impact phase. Similarly, without some means of measuring the direction of the force tangential to the ground surface, the point of mid-stance is ill-defined.

As the horse braces for impact, first the flexor tendons and then the extensor tendons pull in preparation. Barnes and Pinder (1974) took a series of *in vivo* measurements of flexor tendon tension and cannon bone strain. The different signals were correlated using signals from a foot switch and a separate calibration trace.

They noted a rapid increase in lateral bone strain (tension) caused by a pull on the flexors to support the fetlock. Immediately following was a compressive strain of the cannon bone as the extensor tendons position the foot for impact. Preparation for impact, according to Rooney (1977a), is of prime importance in the functional morphology of the leg.

Upon impact, force plate readings typically show the rapid rise in vertical force associated with the heel strike. Pratt and O'Connor (1976) modeled the force trace with the sinusoidal function

$$F(t) = F_{max}\sin(\omega_n t) \tag{4.1}$$

where $F_{max} = 539kg$ was the peak force and $\omega_n = 12.08 rad/sec$ was the frequency of application of the measured force. This approximation holds for the first half of the support phase with the notable exception of a transient form present at the beginning of this phase.

The force trace could then be seen to be a superposition of a smoothly applied vertical force attributed to normal weight bearing and an oscillating vertical force attributed to impact. Pratt and O'Connor then presented a model for that signal as well. Not totally unexpected, the transient signal can be well approximated by the response of an underdamped spring-mass-damper system. The conditions being

$$\omega_n = 186 rad/sec \qquad F(t) = \begin{cases} 52kg & 0 <= t <= .02sec \\ 0 & t > .02sec \end{cases}$$

$$\xi = .23 \tag{4.2}$$

Many other researchers report on similar readings from force plates: Bartel et al. (1978), Kingsbury et al. (1978), Merkens et al. (1985), Rooney (1977b), and Schamhardt and Merkens (1987). Normalization of the time and force are achieved using total stance time and peak reaction force. This normalization turns out to be quite useful in comparing the force traces produced by different horses.

Not all of the results show the occurrence of an impact spike. The use of rubber mats to prevent slipping is one of the reasons. The force plate is sometimes covered by a small thickness of the surrounding ground surface as a means of disguise. This helps to prevent any unnatural gait effects produced by hesitation or fear. Some researchers mention the use of linear extrapolation to remove the transient spike from the otherwise smooth trace.

There are several consistent features of force traces as seen in the literature. Both the vertical and horizontal forces resemble sine functions. If the normalized stance time is of period τ, the natural frequency would be ω_n. The vertical force would act as $sin(\omega_n t/2\tau)$ as it rises from zero to a maximum just before mid-stance. The horizontal force acts as $sin(\omega_n t/\tau)$ with the change from braking to propulsive force at $t \approx (\tau/2)$ being mid-stance by definition.

Since the vertical and horizontal forces can be roughly approximated by the circular sine function, Rooney (1977b) conceptualizes the forces as rotating vectors. Although the resolution of force plate equipment is at its lowest in the impact region, it is a reasonable approximation to consider the resultant force vector at impact to be parallel to the bone column of the first, second and third phalanxes. It is for this reason that the methodology presented in this study uses the natural angle of the horse as the measurement axis.

5. Accelerometric Measurement

A. Measurement Theory

The mechanical elements of any vibration measuring device can be modeled in simplest terms using a spring-mass-damper system (Fig.4). If the system undergoes harmonic motion of the base, and if the principal coordinate $z(t)$ is taken to be the relative motion between the motion of the mass $x(t)$ and the motion of the base $y(t)$, then the steady-state solution of the equations of motion is given by

$$\omega_n^2 z(t) = \frac{1}{\left((1-r^2)^2 + (2\xi\ r)^2 \right)^{.5}} \left\{ Y\omega^2 sin(\omega t\text{-}\phi) \right\} \tag{5.1}$$

where the motion of the base is $y(t) = Y sin(\omega t)$, $r = \frac{\omega}{\omega_n}$, and $\xi = \frac{c}{2m\omega_n}$.

If the phase lag is neglected for the time being, it is easily seen that when the rational quantity is close to unity,

$$\frac{1}{\left((1-r^2)^2 + (2\xi\ r)^2 \right)^{.5}} \approx 1 \tag{5.2}$$

then the term $\omega_n^2 z(t)$ gives the acceleration of the base. This condition is easily satisfied by $\omega_n >> \omega$. This makes $r << 1$ and is the case when the system has a low mass and a stiff spring. The phase lag ϕ turns out to be a constant for cases where the frequency to be measured is smaller than ω_n. This means that the time offset between input and output signals are independent of the frequency of the input signal.

It is easily seen that in order to ensure that the output from an accelerometer is directly proportional to the input and that phase distortion will not be present, the experimental setup must be such that $\omega_n >> \omega$. A test for this condition will be described in a later section.

B. Instrumentation

The PCB accelerometers used in this series of experiments make use of the piezoelectric effect. The spring, mass, and damper elements described in the preceding section are replaced with two polarized quartz crystals, a small seismic mass, and some damping fluid (Fig. 5). When the force from the accelerating mass acts on the crystals, the crystals are strained and there results an electrostatic charge which is proportional to that force. The crystal thus acts as both the spring and the sensor. The polarization in the crystal is then calibrated by the manufacturer, in this case it is roughly $10mv/g$.

The small mass and stiff spring produce a high natural frequency, $\omega_n \approx 100kHz$. This ensures that the theoretical conditions developed above will be met for all but the most extreme cases. There is an experimental condition to be satisfied as well. Any means of attaching an accelerometer to the object of interest by use of an itermediate mounting device or external straps can effect the measurements in terms of frequency response. A later section is devoted to a frequency response calibration of the installed accelerometer and hoof mount.

The hoof mount was forged from aluminum to conform to the shape of the hoof wall. Four holes were drilled through the mount for attachment screws such that each screw was inserted normal to the hoof wall. The site for the accelerometer itself was drilled and tapped according to manufacturer's specifications. The hole was aligned such that the installed accelerometer would be parallel with the hoof wall, with its base approximately $2mm$ from the coronary band.

The accelerometer cable was secured by elastic bandages and tape as it passed up the horse's leg to the instrumentation pack worn around the rider's waist. The rider's pack carried the accelerometer power unit and a Teac seven channel portable data recorder. Foam rubber was used to protect these devices during the testing and also to act as a vibration isolator. The tape recorder was used to capture the signals from either one, two, or three accelerometers and also a voice channel. The use of the voice channel proved to be extremely useful in the identification of the temporal features of the recorded gait patterns.

Once on analog tape, the impact signals were converted to digital form. The replay device was a Teac seven channel data recorder and playback unit with a useful range of gain, calibration, and tape speed cababilities. Analog to digital conversion was performed by a Zenith 150 micro-computer and a Metrbyte Dash-16 DMA board which provided twelve bit resolution. Data was converted at $1000\,Hz.$ and stored on floppy diskettes. The data files typically contained 9000 data points or about $9\,sec.$ of continuous information. At a collected gallop, this amounted to 16-17 strides per data file.

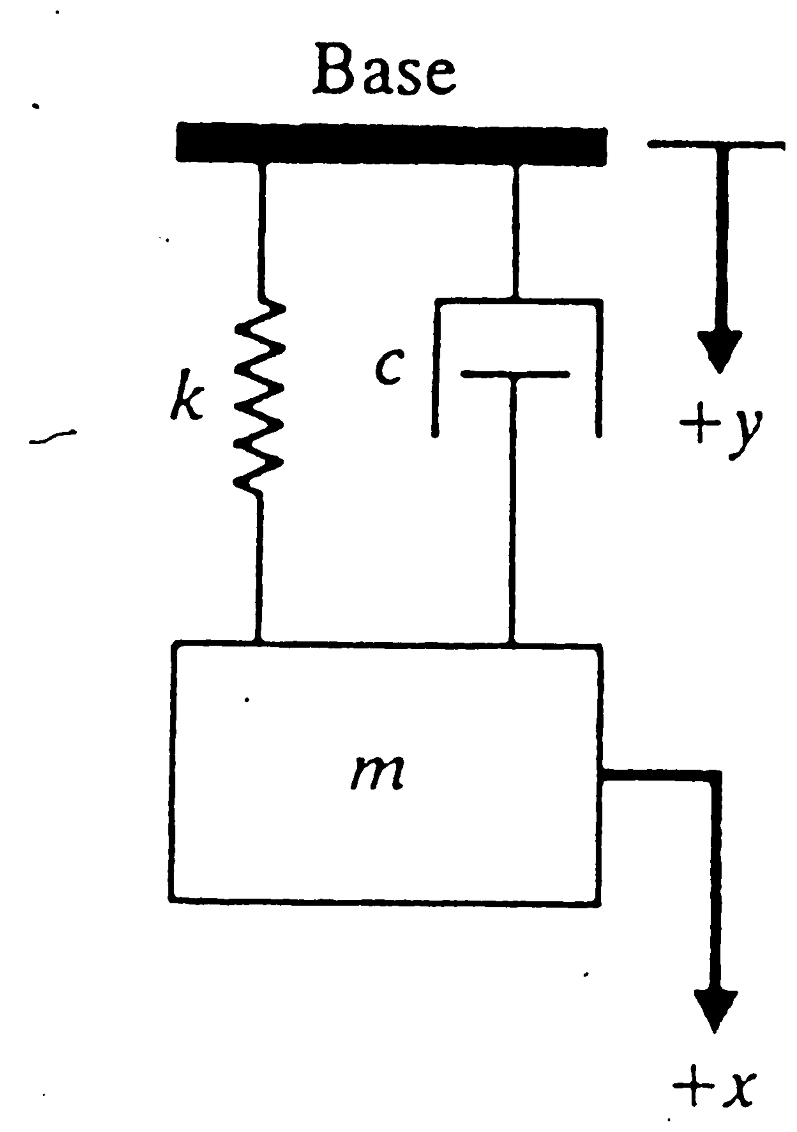

Fig. 4. Spring-mass-damper system with harmonic motion of the base.

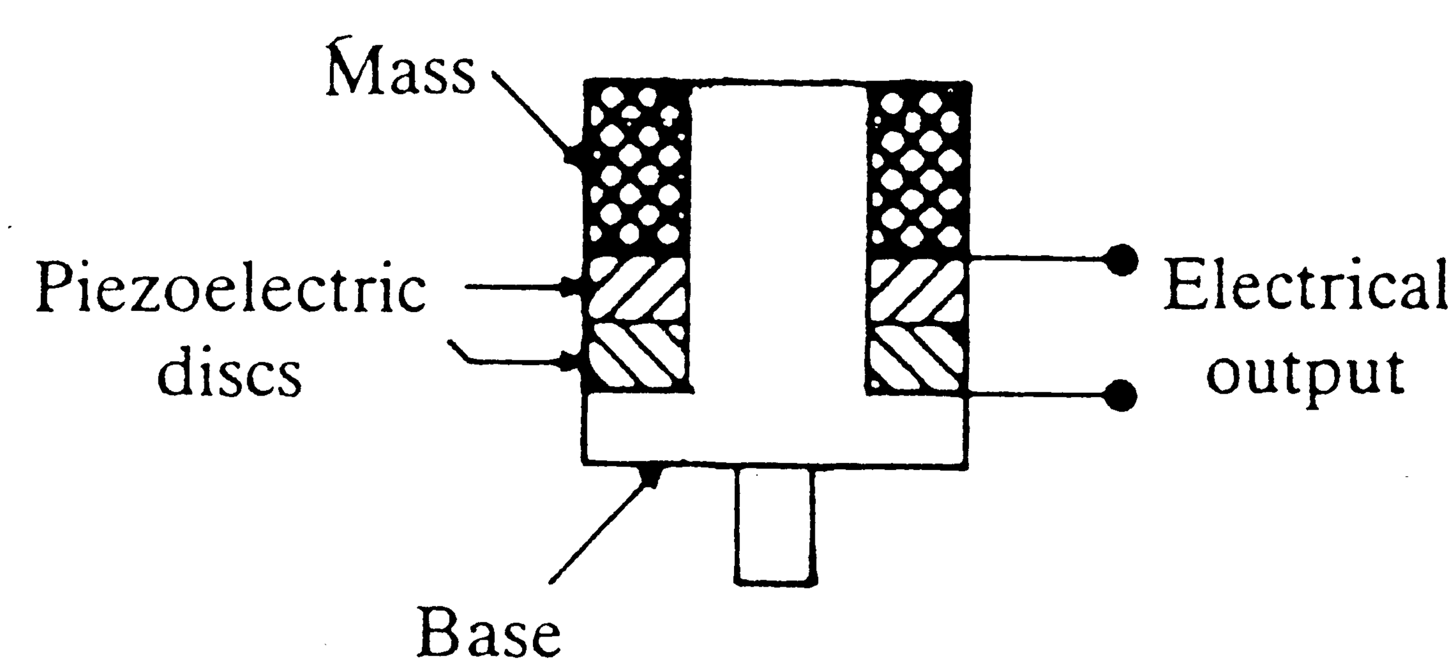

Fig. 5. Schematic diagram of the accelerometer.

6. Verification of Surface Mount Technique

This technique of verification provides comparison between two experimental methods in analyzing the shock wave which passes through the third metacarpal under an impact load. The first method involves collecting strain measurements during impact. The second set of data points come from an accelerometer which is mounted on the bone at approximately the same location along the axis of displacement as the guage. The two sets of temporal measurements can be compared after either a double differentiation of the strain or a double integration of acceleration.

Since numerical differentiation of discrete functions is notoriously unstable and since this study would require differentiating strain twice with respect to time, it was decided to compare the two measurement techniques in terms of displacement.

Clean, dry bone samples were obtained through the cooperation of the Comparative Orthopaedic Biomechanics Laboratory at the University of Pennsylvania Veterinary College, New Bolton Center. Only one bone was used in this study and it was prepared in the following manner.

The installation site for the guage was filed flat, cleaned and neutralized. A 120Ω foil strain guage and soldering tabs were glued to the specimen using cyanoacrylate and the electrical connections were completed. Upon checking the resistance of the circuit, the complete installation was sealed. To hold the accelerometer, a small steel mount was drilled and tapped. Next, the site for the accelerometer mount was cleaned and neutralized and the mount was glued to the site with epoxy.

A check on the completed installation showed that after an initial disturbance, the metacarpal bone would resonate if not damped. If the bone was held, as it was during the rest of the experiment, little if any resonance was observed. This is consistent with the notion that the equine forearm musculature serves to damp vibrations.

The guage was wired as an arm of a quarter-bridge circuit and the output of the strain indicator was monitored on a digital oscilloscope. The strain indicator had a variable gain output which was calibrated using static loads. The calibration procedure entailed application of loads to the bone and plotting strain as read from the indicator vs. voltage as measured on the oscilloscope. The correspondence between the two was almost perfectly linear (correlation coefficient = .997) with slope of roughly $2\mu\epsilon/mv$. This value was used in the data reduction process to convert digital data values to bone strain values.

During the experiment, the strain indicator signal was monitored on the scope and simultaneously recorded on the analog tape recorder with a gain of 5 to further enhance resolution. The signal from the accelerometer was likewise monitored and recorded but with no gain. Impact loads were achieved using a simple drop hammer device and several impacts were recorded.

The analog recordings were then stored in digital form using a sampling rate of

$10KHz$. If some sort of timing device was used to start the A/D conversion at the precise instant before impact, better advantage could be made of the hardware's $25KHz$ maximum sampling rate. As it was, the 9000 data point maximum was the limiting factor and $10KHz$ was the maximum sampling rate in terms of being able to fully bracket the impact.

A typical pair of acceleration and strain plots are shown in Figs. 6 and 7. Extrapolation of the strain guage results is staightforward if it is first assumed that the strain measured by the guage at the midpoint to represent the average strain of the entire bone under compression. The second assumption is that the cross section and modulus of the bone remain constant. The size of the guage is well above the minimum useful size recommended by Barnes and Pinder (1974). This, together with the fact that the wavelength of the shock is of $\mathcal{O}(10^2)$ times the length of the bone justifies the assumptions as first approximations.

In this manner, multiplying the average strain of the column and the length of the column below the point of measurement gives the displacement of that point. Performing this operation for the strain measurements results in a discrete displacemnent function tabulated at $.0001sec$ intervals.

To reduce the acceleration measurements to displacement, several methods were tried with varying results. Since the temporal acceleration data is a discrete function with equally spaced abscissas, the obvious integration procedure will in some way involve adding up the value of the integrand at a sequence of time steps within the range of integration. Of course, there are choices of different methods of various orders but the best starting point is usually with the trapezoid rule.

The trapezoid rule for a function defined at equal time intervals is given by

$$\int_a^b f(x)\,dx = \mathrm{h}\Big(\tfrac{1}{2}f(a) + \tfrac{1}{2}f(b)\Big) + \mathcal{O}(\mathrm{h}^2 f'') \tag{6.1}$$

where the error term $\mathcal{O}()$ is an estimate of the error in taking one step. This rule is only exact for polynomials up to and including degree one and it was tempting to try Simpson's rule. Integration then would benefit by a cancellation of terms in the Taylor series expansion of Simpson's rule and make this formula exact for polynomials up to and including degree three.

It turns out that this happens to be one of the cases where higher order does not mean higher accuracy. Since there is a fixed number of data points, using Simpson's rule effectively cuts the number of intervals of integration in half. By using cubic splines as a continuous approximation to the discrete function, it would be possible to integrate using smaller stepsizes or even use a program with adaptive stepsize. But the present problem is just to check on the agreement between the two experimental methods and the trapezoid rule seems to serve well.

Another method of obtaining displacement from the accelerometer data uses the central-difference approximation to the second derivative of a function:

$$y_i'' = \frac{y_{i+1} - 2y_i + y_{i-1}}{h^2} \tag{6.2}$$

Writing this equation for n sampling points results in a system of n+2 simultaneous equations which to be solved for the unknown y_i. By introducing the constraints that displacement is zero at some time before and at some time after the impact, two more equations can be introduced. The complete system is tridiagonal and easily sovled.

Figure 8 displays the displacements of the measurement point as found by using three different methods: trapezoid rule numerical integration, boundary value problem formulated by finite differencing, and extrapolated strain guage data. The error in calculated peak displacement are 21% for the trapezoid integration. Finite differencing produced only 13% error at the peak and more closely resembled the strain data.

Two conclusions can be reached from the close agreement in form between the results from the strain guage and those from the accelerometer. The strain guage instrumentation used here can be considered so reliable that the results gained from analysis of the dynamic strain can be used as a benchmark. The first conclusion, then, is that externally fixing an accelerometer to the horse will produce meaningful measurements of impulsive shock waves transversing the point of attachment.

Secondly, measurements of acceleration can be used for comparison with measurements of strain. For example, if multiple accelerometers were located along the horse's leg estimations of bone strain could be made without surgical implantation of strain guages and subsequent destruction of the animal. Several accelerometers would be needed to allow for the separation of rigid body motion from strain deformations.

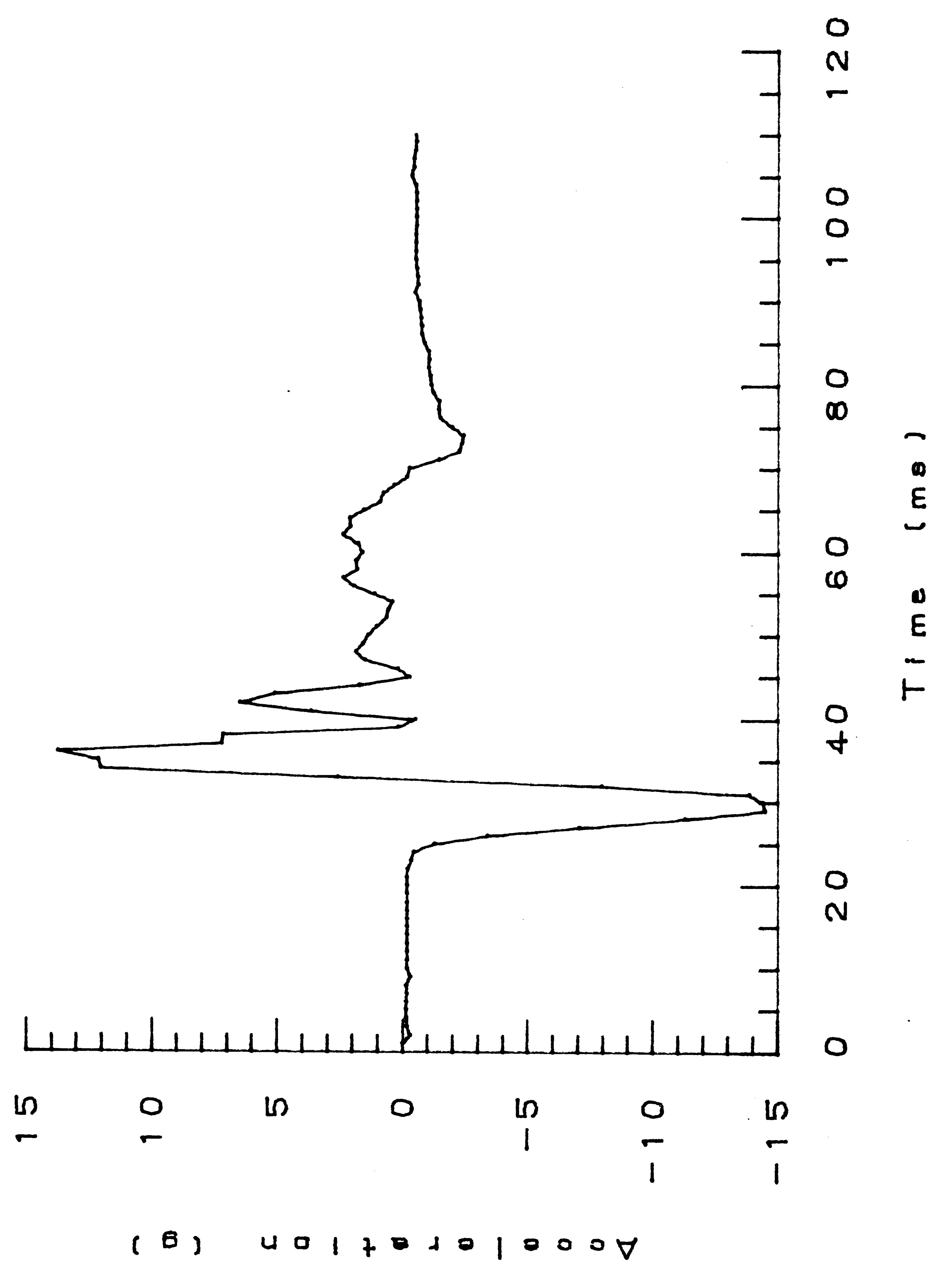

Fig. 6. Accelerations measured from the third metacarpal under an impact load.

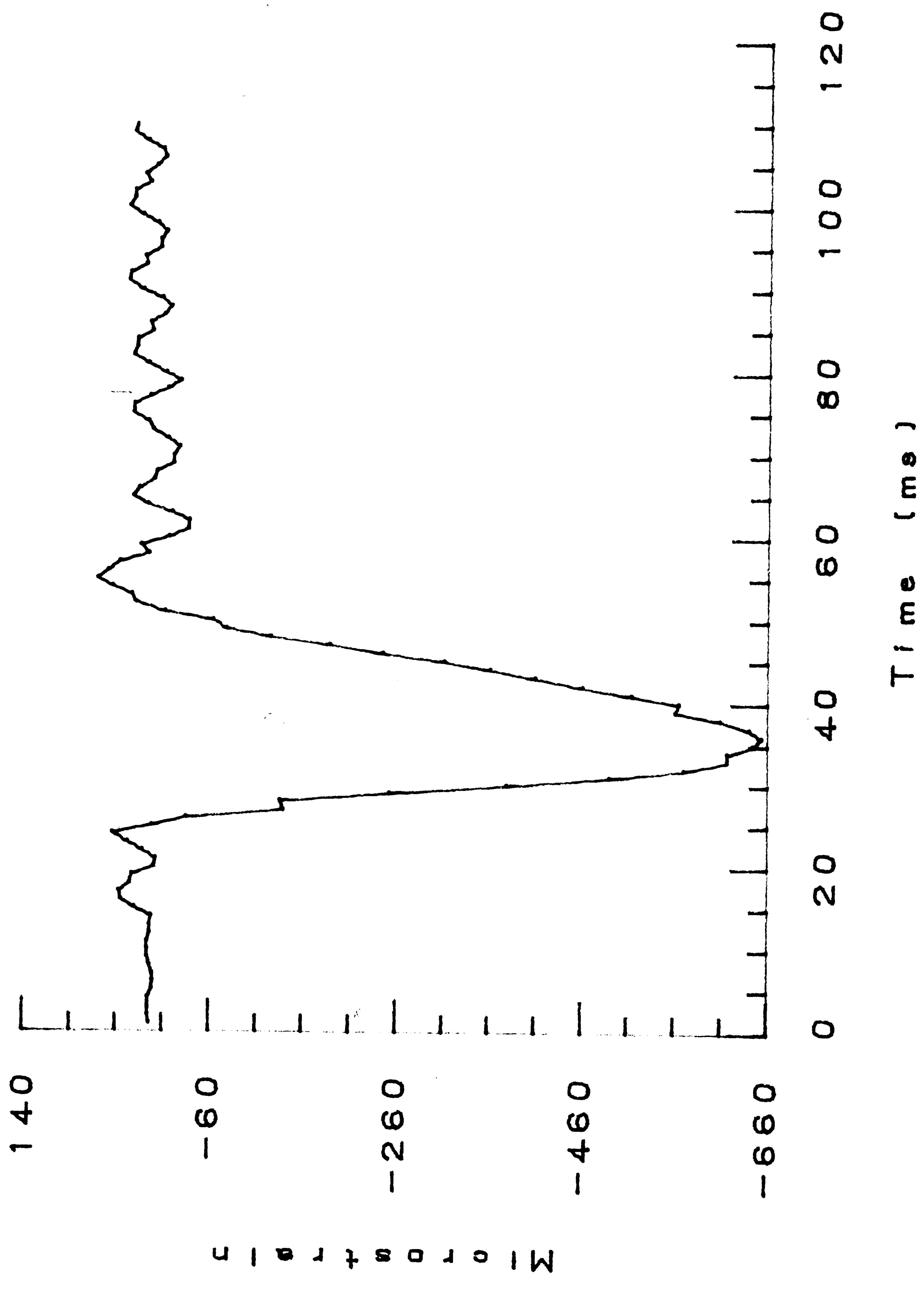

Fig. 7. Strain measured from the third metacarpal under an impact load.

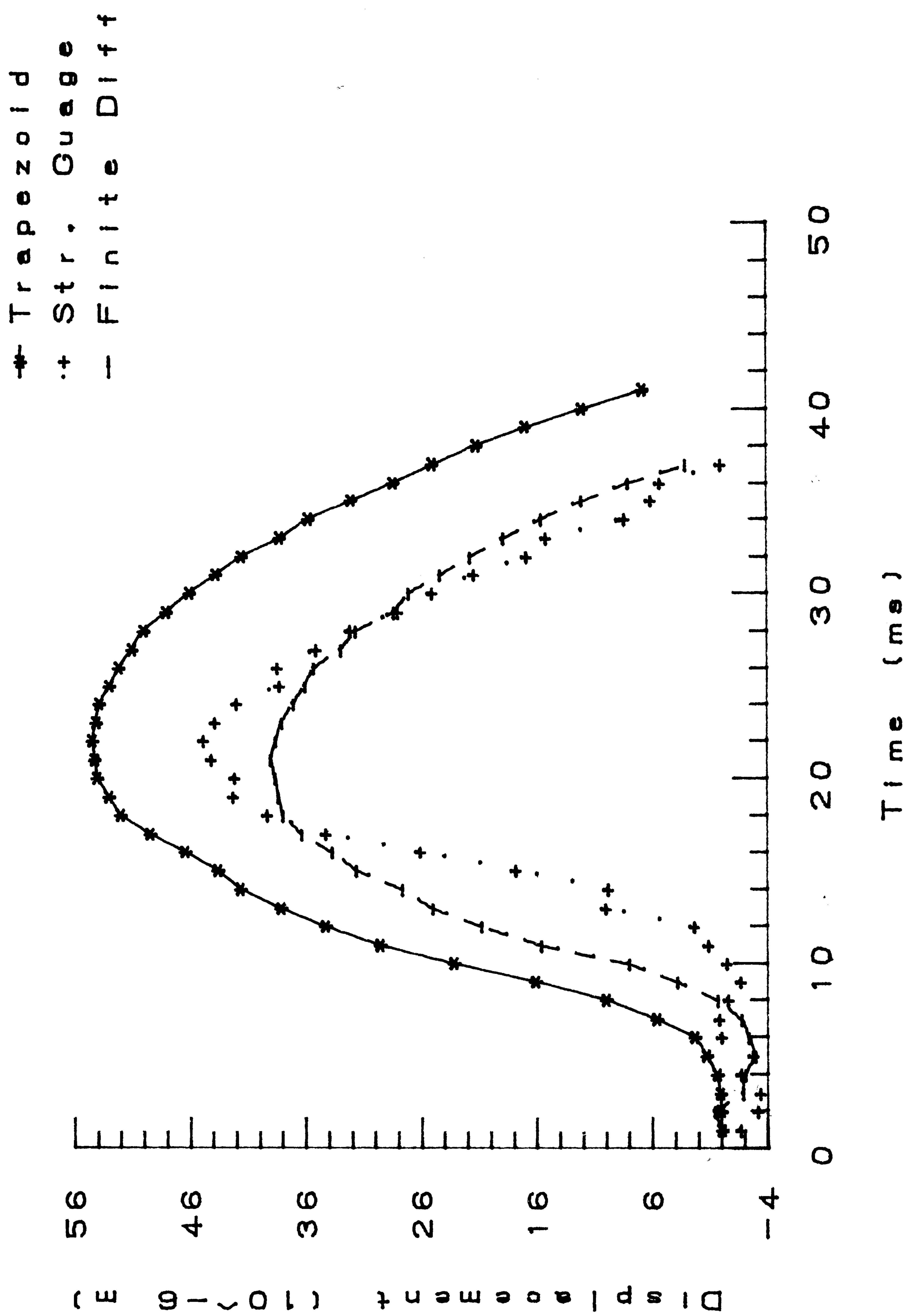

Fig. 8. Three calculations of the displacement of the midpoint of the third metacarpal.

7. Data Collection Procedures

Choosing the proper horse to be the test subject is an important consideration. For this type of testing, the horse must be sound, quiet, have steady gaits and healthy hooves. The subject in each of the tests described in this paper was Sweet Sandy Dandy, a nine-year-old Quarter Horse gelding from solid bloodlines. He has proper conformation and is not predisposed to any lameness. Careful attention was given to his training such that the horse would not become fatigued from any of the longer tests.

Sandy possesses a calm disposition and required only one session to become used to the wiring procedure and apparatus. He would stand for shoeing or being wired without additional help and at no time fussed over or tired to bite at the accelerometer cables. Another advantage of this horse over others considered for testing was the state of his hooves. His hoof wall is thick and solid and served quite well in holding the accelerometer mount in place. When properly maintained, his hooves were observed to grow at a rate of roughly $1cm$ every three weeks. Since some of the tests involved changing shoes several times and since each test required another set of screw holes from the mounting plate, this rate of growth was desireable.

Each test was preceded by a careful job of shoeing. Usually done two or three days before the test, aluminum outer rim shoes were selected to provide reliable traction over a range of surfaces. This job was in each case done by a professional racetrack horseshoer and special attention was given to be sure that the horse was balanced and that the hoof flight patterns were proper.

The first step in setting up for a test was for the horseshoer to affix the accelerometer mount to the hoof wall. Holes were layed out with a template and drilled to the proper depth. Next, the mounting plate was fastened to the hoof and the accelerometer screwed into the mount (Plate 1). The cable was then run from the accelerometer up the horse's leg and taped in place with strain relief loops as suggested by the manufacturer (Plate 2). First elastic tape and then bandages cover the installation. Polo bandages were used on the leg and Vetrap bandaging tape was used on the hoof. The tape and polo bandages worked well in keeping the cable secure and protected. The Vetrap also served to keep moisture away from the accelerometer-cable connection. Plate 3 shows the completed set up with the rider mounted. Plate 4 shows data collection in progress.

Plate 1. View of Accelerometer Attachment.

Plate 2. View of Left Foreleg with Cable in Place

Plate 3. Completed Installation.

Plate 4. Use of the System for Data Collection.

8. Natural Frequency of Mounted Accelerometer

The natural frequency of the instrumented hoof was calculated as a part of a larger series of tests which was performed during the same afternoon. Recordings of impact signals were made using a dirt surface at the walk, trot and gallop. These recordings will be discussed in the following section and showed no signs of improper instrumentation or data collection procedures. The test horse was then galloped over a macadam surface, a set of conditions which induced resonance of the instrumentation.

These resonant signals were characterized by magnitude and frequency which were disproportionately high as compared to a gallop on the dirt surface or a trot on the macadam surface. No gait on the dirt surface and none of the slower gaits on the macadam surface produced impacts which induced anything which resembled resonance. When the spectral density of the resonant signals was investigated, it showed to have a strong band centered at $\omega \approx 300 Hz$.

This value is taken to be the frequency of the fundamental mode of the instrumented hoof. In fact, since the structure of the hoof is strongly damped and since the accelerometer has a natural frequency of $\omega_n = 100 kHz$, what was observed was either the fundamental mode of the accelerometer mount or the mounting screws. Exactly which component resonates is of little concern, it is the magnitude of the lowest natural mode of the system which is important. If this value is too large to satisfy equation 5.2, then our measuring device cannot be used for investigation. With ω_n=300 the measurement error will be less than 10% for all cases of interest.

9. Patterns of the Natural Gaits

The testing procedure previously described was used many times, over a wide range of conditions. In sections 10, 11, and 12, tests will be discussed which included the variation of gait pattern, ground surface material, ground surface condition, and shoeing techniques. This section presents some typical results for each of the natural gaits, and the results shown here will form a basis for further comparison.

A. Time Domain Patterns

For the purposes of these tests, four different gait patterns will be discussed: foreleg at a walk, foreleg at a trot, non-lead foreleg a a transverse gallop, and lead foreleg at a transverse gallop. Plots of the time domain patterns are shown in Figs. 10-13. Plots of the spectral density of the impact phase of these gait patterns are shown in Figs. 14-17. The ground surface used in obtaining the data shown in Figs. 10-17 was a riding ring with a dirt surface. Several tests were performed on this surface over a period of three months with excellent repeatability.

In Figs. 10-13 there are two sharp, well defined peaks of positive acceleration which were found to be consistent features in the first recordings. Obviously, one is impact and the other is take-off but the making the distinction was difficult. Using the voice track to record the rider's observation of the footfall cadence worked fine for the walk and trot but was inadequate for the gallop. For the purpose of defining the features of the accelerometric patterns, a special test was devised.

The horse's hind shoes were removed and the microphone was attached to the breastplate and pointed downwards. With the testing surface changed to asphalt, the recording system had a built-in trigger, namely the sound of the horse's shoes striking the ground. The data was replayed to an oscilloscope with the playback unit tape speed set to one fourth of the recording speed. Separation of the features of the data was then a simple task. In each of Figs. 10-13 the first peak is impact and the second is take-off.

Besides the impact and take-off peaks, there are other consistent similarities between the four patterns which are noteworthy. Just prior to impact there exists a period of relatively low frequency, positive acceleration. This is suggested to be due to the bracing, or pretensioning of the leg in readiness for impact. The point of impact can be well defined by a close look at the individual data points.

In every case, it is possible to identify a boundary between the low frequency pre-impact phase and the rapidly changing, higher frequency impact phase. To illustrate, a period of $40ms$ which contains the point of impact of Fig. 13 has been magnified in Fig. 14. In the magnification, the first point shown is τ=.183sec and the last τ=.223sec. The point of impact is defined to be τ=.203sec as indicated by the marker drawn on Fig. 14.

The accelerations at take-off also show consistencies between the four gait patterns. Comparing the lead foreleg and non-lead foreleg at the gallop (Figs. 12 and 13 respectively) it is evident that even though the peak impact accelerations differ ($15.4g$ vs $23.1g$), the peak take-off accelerations are the same ($25.7g$) for similar stride durations (.551*sec* and .556*sec*).

A summary of peak accelerations and stride duration of the single typical strides shown in Figs. 10-13 is presented in Fig. 9 below. The fact that peak take-off accelerations are greater than peak impact accelerations is not something that would be intuitively obvious. This may be due to the fact that at impact, the hoof is moving slower than the center of mass of the horse while immediately after take-off the hoof is moving faster than the center of mass. The hoof thus undergoes a greater total change in velocity at take-off than at impact. Since the area under the acceleration curve is equal to the change in velocity, it is reasonable that the take-off peak is of greater magnitude than the impact peak.

This occurrence has been documented in the literature. Pratt (1985) discussed similar accelerometric recordings obtained from the hoof of a galloping horse. In that study, the measurement axis was perpendicular to the axis used here but take-off accelerations were still higher than those of impact. Barnes and Pinder (1974) noted that most of the large and rapid changes of bone strain of the horse at the walk occured during the swing phase rather than the impact phase.

Fig. 9. Summary of Temporal Pattern Features of the Single Strides in Figs. 10-14.

Figure	Gait	Peak Acceleration Impact	Peak Acceleration Take-off	Stride Duration
Fig. 10	Foreleg at a Walk	$8.1g$	$11.6g$	$1.140sec$
Fig. 11	Foreleg at a Trot	$10.2g$	$14.6g$	$.705sec$
Fig. 12	Lead Foreleg at a Gallop	$15.4g$	$25.7g$	$.551sec$
Fig. 13	Non-lead Foreleg at a Gallop	$23.1g$	$25.7g$	$.556sec$

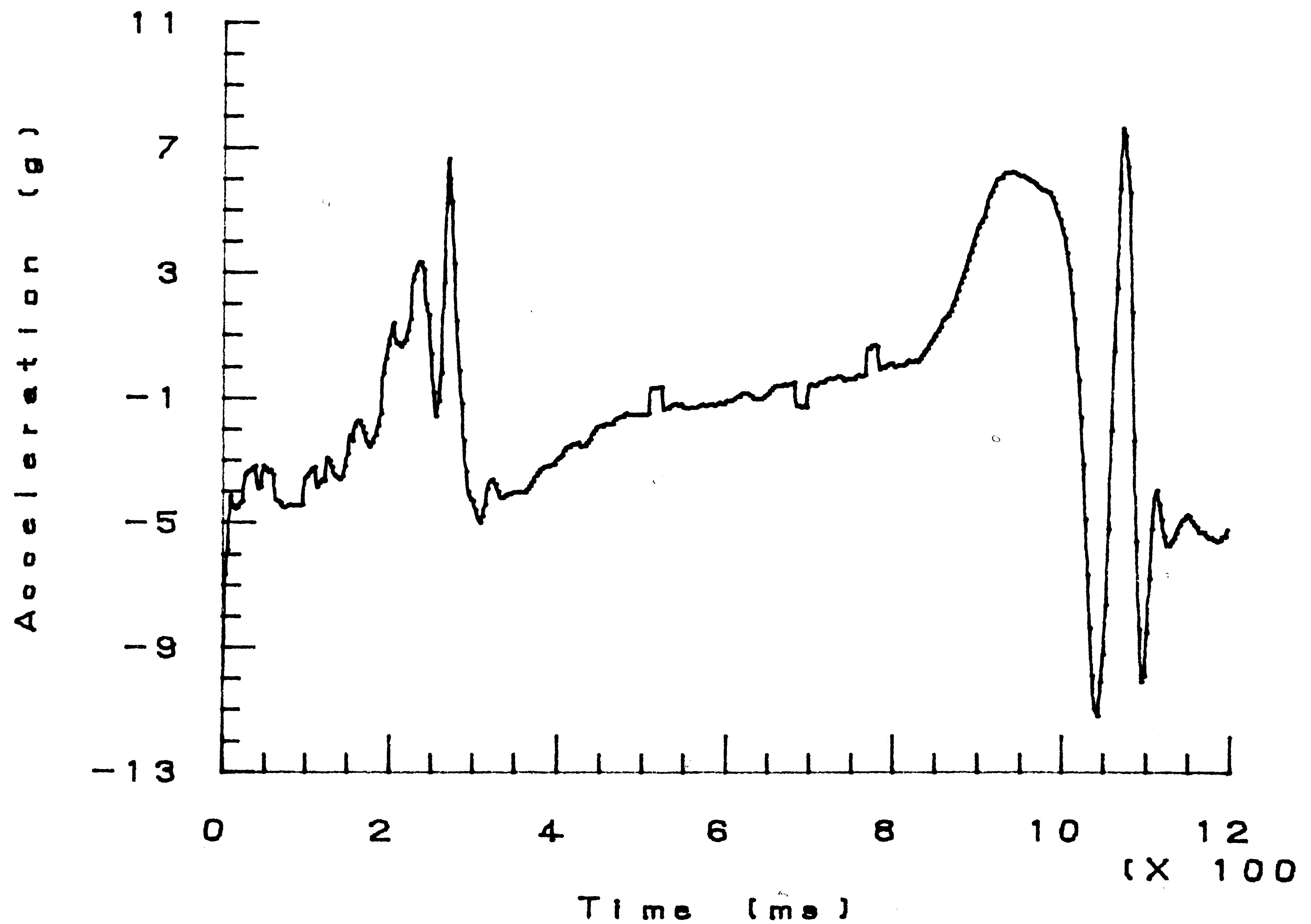

Fig. 10. Temporal gait pattern: Foreleg at a walk.

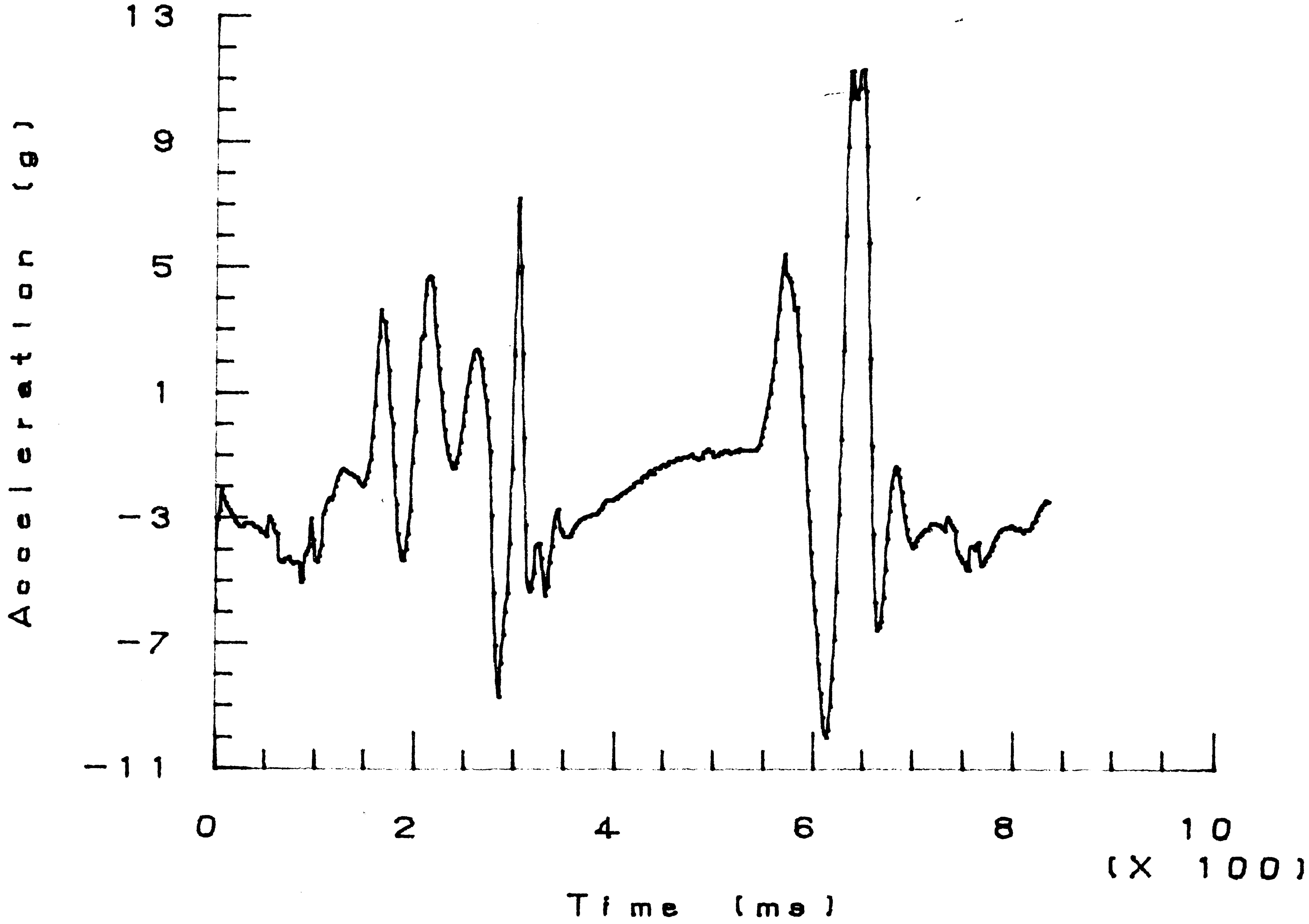

Fig. 11. Temporal gait pattern: Foreleg at a trot.

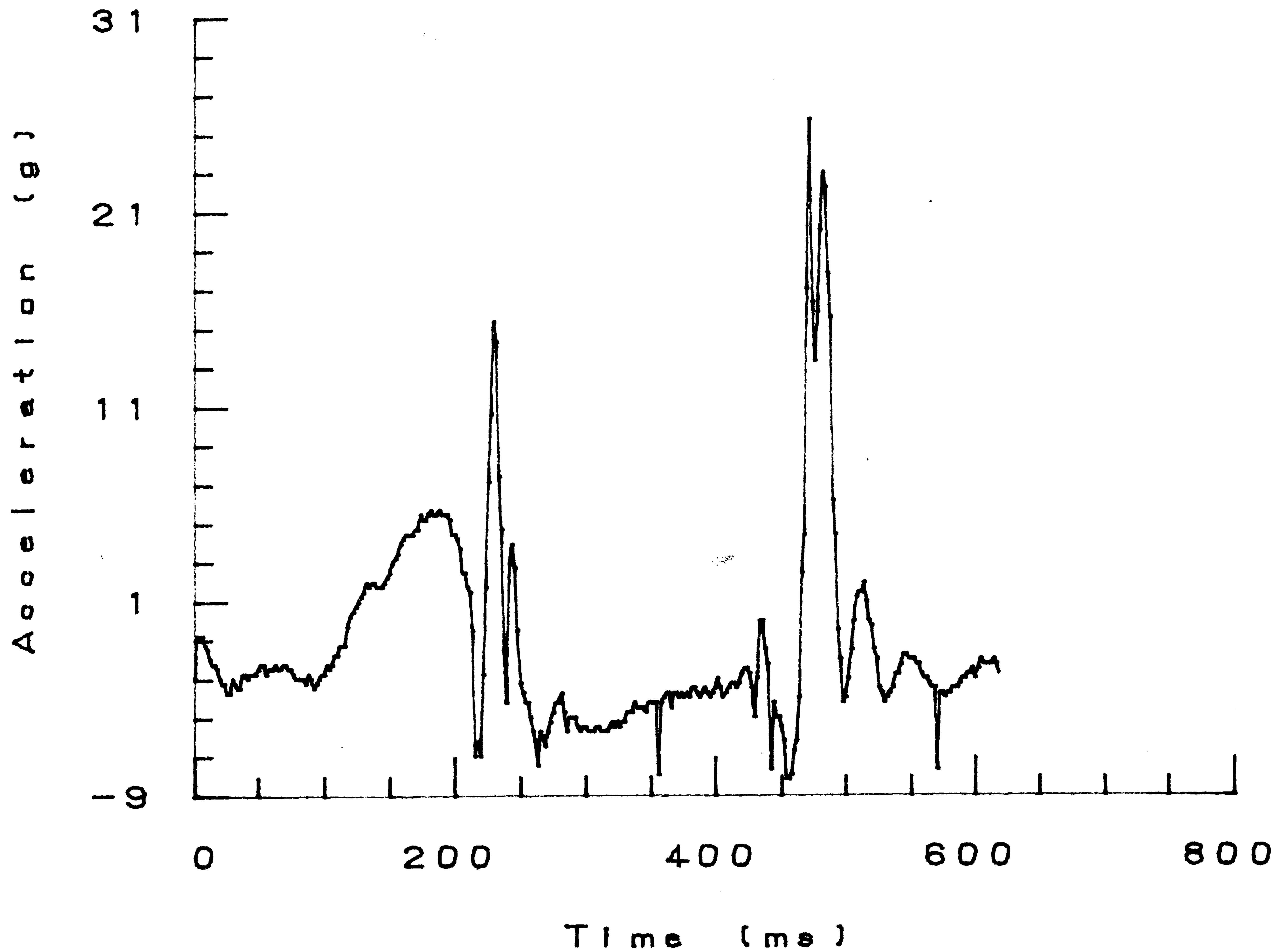

Fig. 12. Temporal gait pattern: Lead foreleg at a gallop.

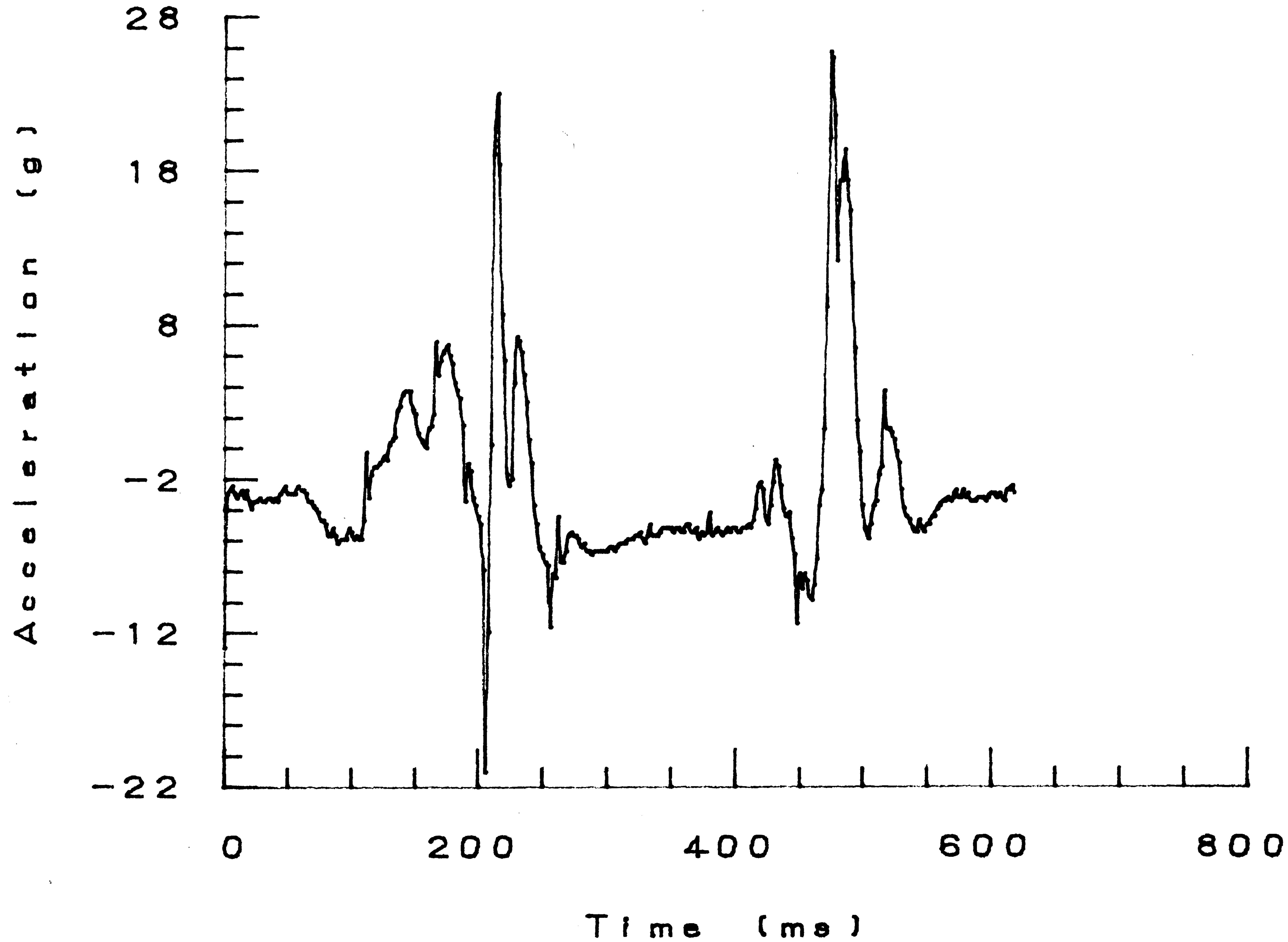

Fig. 13. Temporal gait pattern: Non-lead foreleg at a gallop.

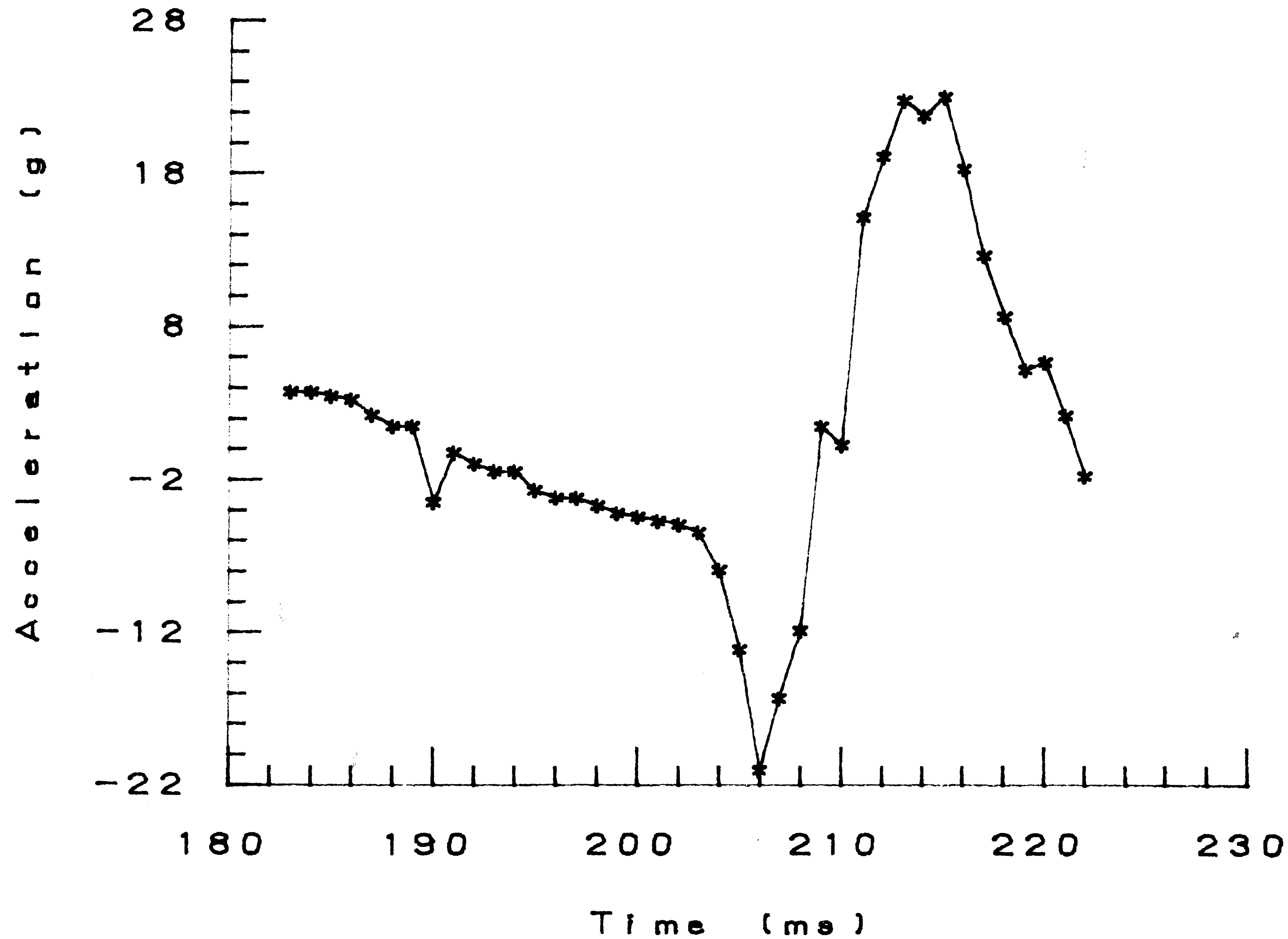

Fig. 14. Point of impact.

B. Frequency Domain

Spectral densities were calculated using an interactive Fast Fourier Transform program written by Scott Wunderlich, 1988. The program has a number of useful routines for massaging data in time and frequency domains including digital filtering, inverse FFT and data windowing. The window routines were used to determine the frequency of the impact shock by centering the window on the impact peak. Windowing was used with a sin^2 function to prevent frequency leakage from the limits of a square window. This procedure made it possible to investigate typical spectral densities of impact for the four gait patterns and under varying conditions.

The spectral density of impact of each of the four natural gait patterns are shown in Figs. 15-18. These discrete functions were obtained from the signals shown in Figs. 10-13. A generalized comparison between the impact signals from the natural gaits can be made. As the speed of the gait progresses from the walk to the trot to the gallop, the magnitude of the peak acceleration in the time domain increases. The frequency domain of the impact behaves in a similar manner as the peak magnitude of the power density function also increases. In addition, the frequency bands shift from lower to higher as the gait goes from walk to gallop.

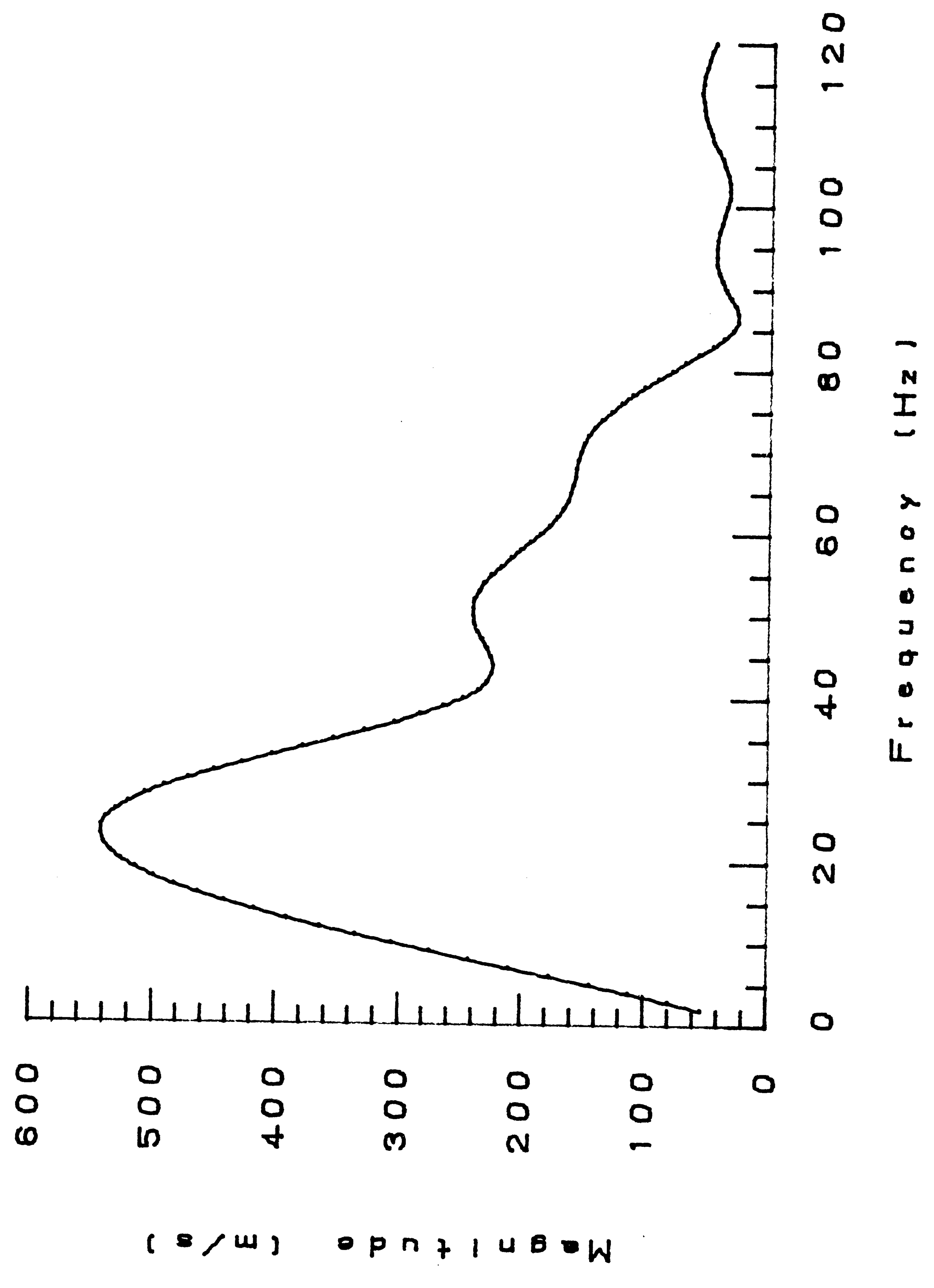

Fig. 15. Spectral density of foreleg impact at a walk.

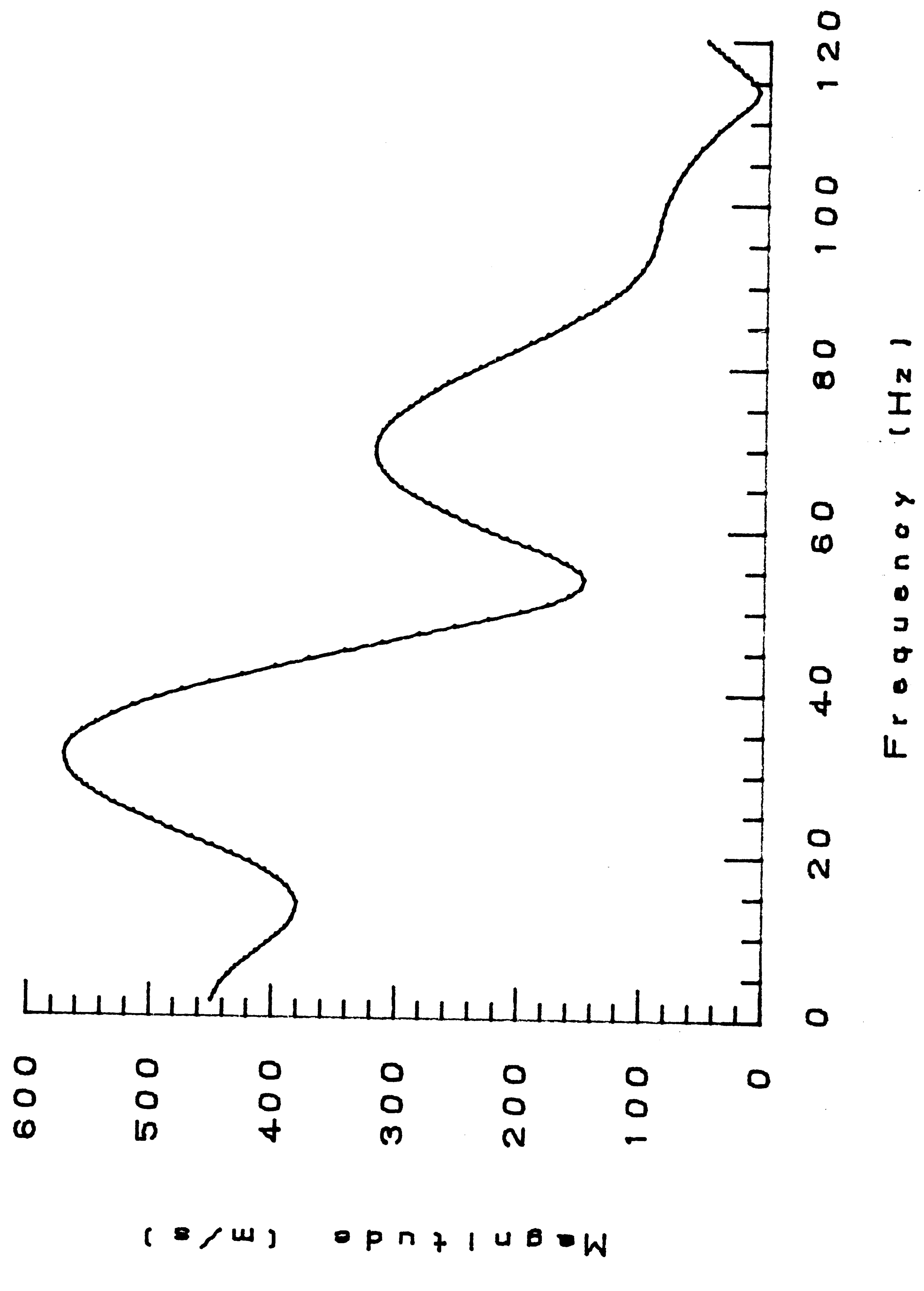

Fig. 16. Spectral density of foreleg impact at a trot.

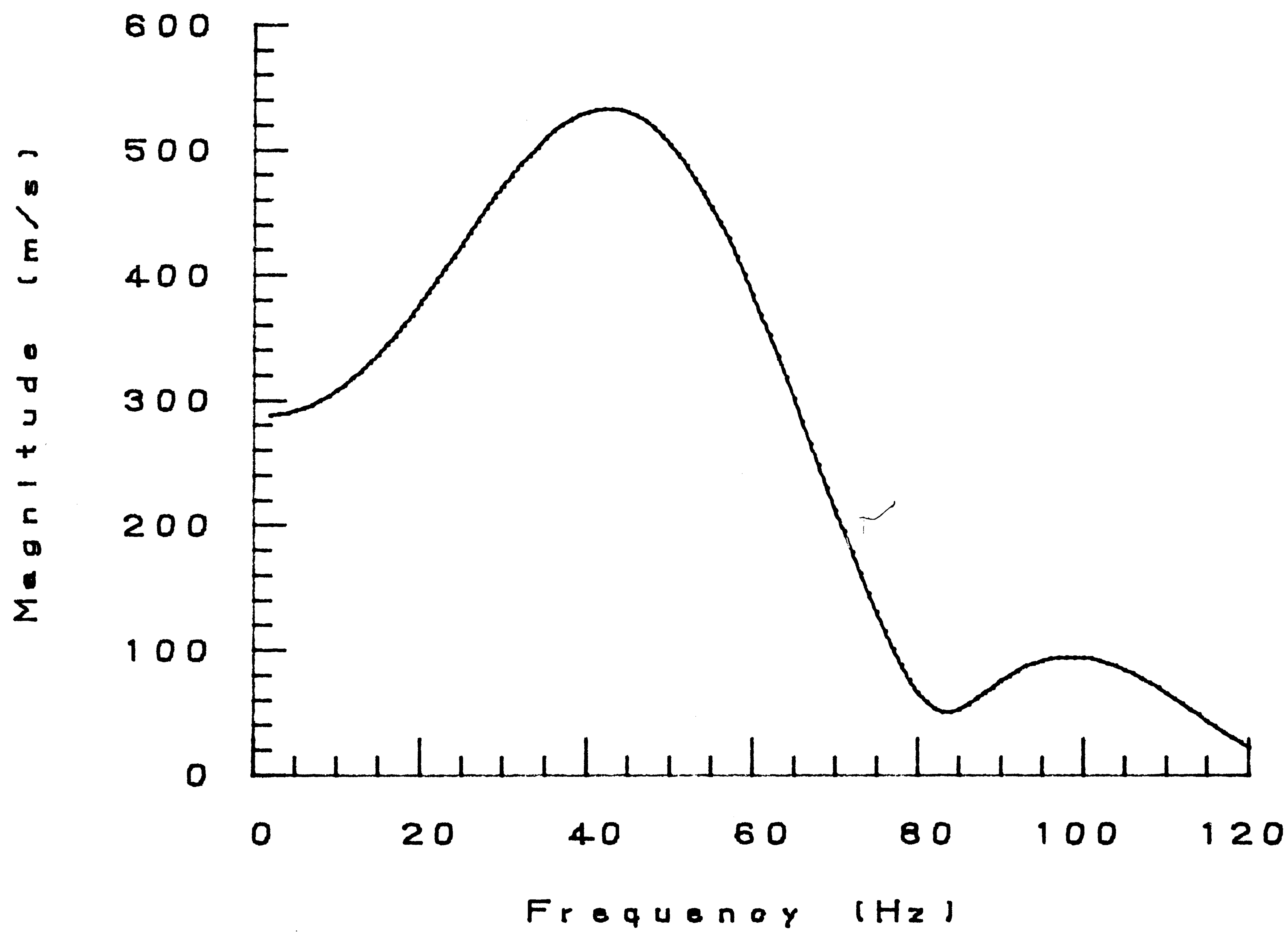

Fig. 17. Spectral density of lead foreleg impact at a gallop.

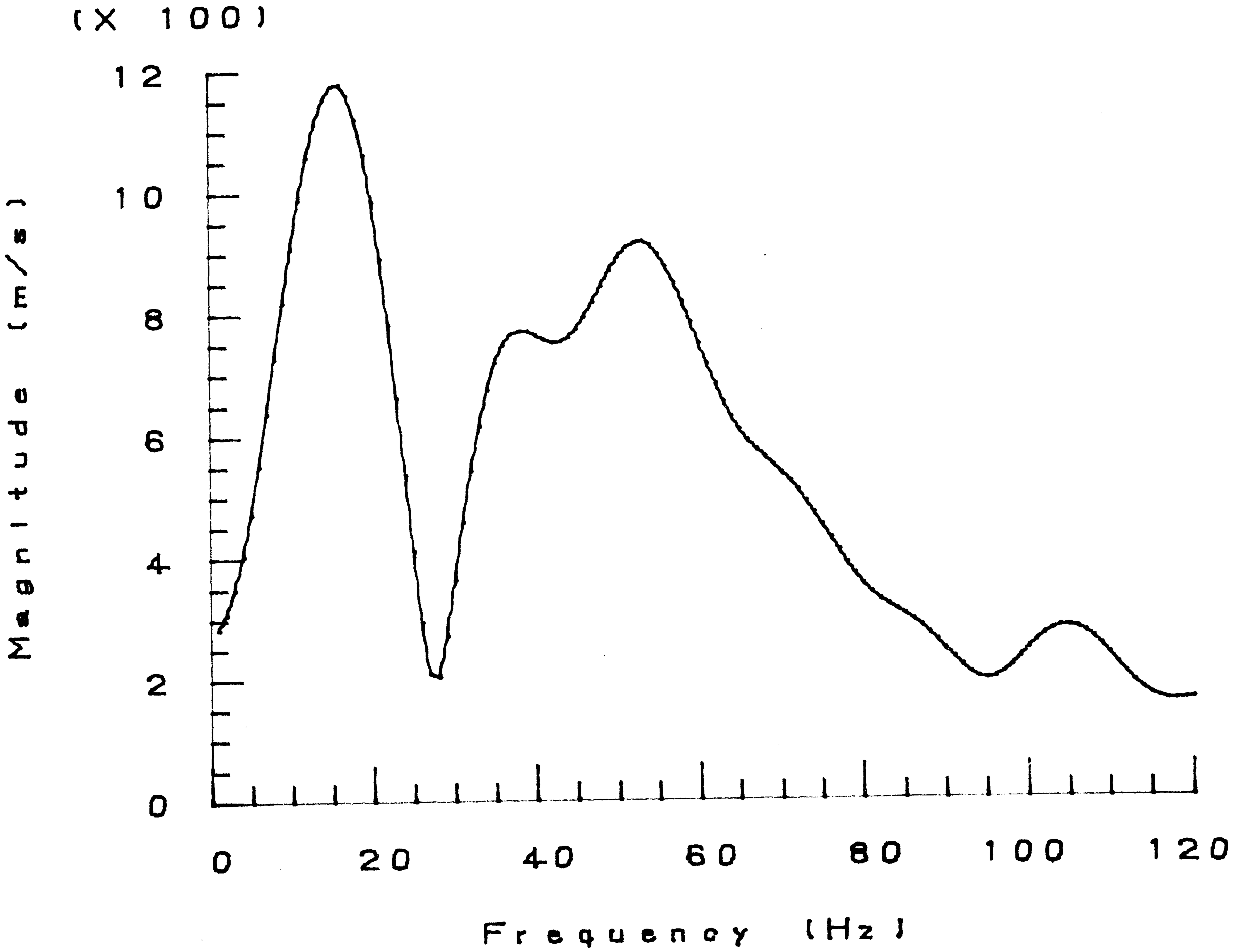

Fig. 18. Spectral density of non-lead foreleg impact at a gallop.

10. Averaging as a Means of Data Reduction

The combination of peak analysis and frequency analysis provides a useful combination of data reduction techniques and will be used in tandem in the presentation of data in this report. While the single strides discussed above are representative of the natural gaits, the following sections will use data averaging from successive strides to characterize the experimental conditions. The number of strides taken to be averaged was always greater than forty.

Peak acceleration analysis was generally performed with an interactive peak finding program written by Scott Wunderlich (1987). By identifying the peak accelerations from consecutive strides and then obtaining the average and standard deviation of those peaks, useful measures of the temporal impact accelerations were obtained. These results will be presented as peak acceleration $(g) \pm$ standard deviation (g).

Statistical testing can then be performed on these groups of peak to determine if there is a significant difference between them. For instance, in the following chapter, we would like to know if the observed difference in the average peak impact accelerations is due to an actual difference between the lead and non-lead legs or if the difference is due to random sampling error.

The average peak acceleration was a useful index for such experiments as that involving shock absorbing horseshoe pads. Standard deviation was a more practicable index for rating the effectiveness of surface maintanence procedures since the standard deviation of peak acceleration is assumed to be related to uniformity of the ground surface. Keeping the surface uniform is a prime concern of racetracks, polo fields and other equine event arenas.

Power density functions were likewise calculated for forty or more contiguous impacts and then averaged harmonic by harmonic. The resulting spectra were similar in form to those of the single strides shown if Figs. 15-18 with the exception that the averaging process tended to widen the frequency bands and lower the peak magnitude. While we don't have the statistical resourses available for time domain analysis, useful comparisons are made through the use of transfer functions.

Transfer functions compare the spectral density of one averaged group of functions to another averaged group by finding the ratio of the two. The transfer function is calculated harmonic by harmonic. Frequency domain analysis was found to be an important means of comparison of different gaits, between different conditions at the same gait, and between the same gaits on different surfaces.

11. Comparison of Impact of the Lead and Non-lead Foreleg at the Gallop

The dependence of the spectral density function and peak acceleration on gait was almost intuitively obvious. Less apparent were the distinctions between the lead and non-lead foreleg at the gallop. The legs of the horse have different functions depending on the sequence in which they are placed on the ground. In general, the horse will lead with the innermost foreleg while turning at the gallop. On long straight stretches, the horse will most likely change leads from time to time in order to avoid fatigue.

If force plate data were available to differentiate between the two gait patterns, it is likely that there would be a difference in the amount of weight borne by the forelegs and in the time to reach peak force. The size of the force plate itself makes it difficult to take measurements of the faster gaits. For example Merkens and Schamhardt (1985) reported an average of 2.9 attempts per data point in recording ground reaction force patterns at the walk. They were understandably particular about data collection and discarded those trials in which oblique footfalls or near-misses were recorded. The problem as it relates to the lead/non-lead question is that while 2.9 attempts per data point is tolerable, it required 20 attempts per acceptable data point at the trot.

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Data collection with on-board recording not only provides enough strides to study, it provides the researcher with information concerning contiguous strides as well. In one series of experiments, a review of over 80 strides on each lead was conducted with the following results. The average peak acceleration for the non-lead leg was $26.4g \pm 6.8g$ while the average for the lead leg was $11.86g \pm 3.8g$. The confidence level for the difference in these means was over 99%. This gallop data was a part of over 600 strides studied under varying conditions from that surface on that day. The ratio of non-lead to lead impact accelerations remained constant for all of those tests with an average of ratio of 2.05.

The percent standard deviation of the galloping peak impact accelerations held constant at about 30%. The standard deviation of peak accelerations did not lessen substantially at other gaits and was the main reason for considering such a large number of impacts. Merkins and Schamhardt (1985) were the only researchers to report on the standard deviation of force plate data. At a walk, they found 3.3-7% standard deviation in vertical force, 10% in linear force, and 20% in lateral force. While these figures are somewhat lower than those obtained here, it must be remembered that Merkins and Schamhardt (1985) selected their data points with care. It may be that such a procedure could be adopted with accelerometric data but no acceptable basis for discarding questionable data points was found.

It was noticed that some portions of the data, say five to ten strides, are quite uniform and exhibit very nearly equal impact accelerations. Other portions of the recorded data are anything but regular and are often characterized by one or more very large peaks followed by one or more very small peaks. If any discriminative data selection procedures were needed, selection of these uniform clusters of impacts would be a reasonable starting point.

The second comparison between the impact of the lead and non-lead foreleg at the gallop is in the frequency domain. First, power density spectra were averaged using 64 strides for each foreleg. These average spectra are shown in Fig. 19. A transfer function was derived using the lead foreleg as the basis, and the non-lead foreleg for comparison. Figure 20 shows a plot of this transfer function and it is evident that except for the region of 1-2Hz, every frequency of impact shock is greater for the non-lead leg. The regions of greatest difference are concentrated in two distinct bands.

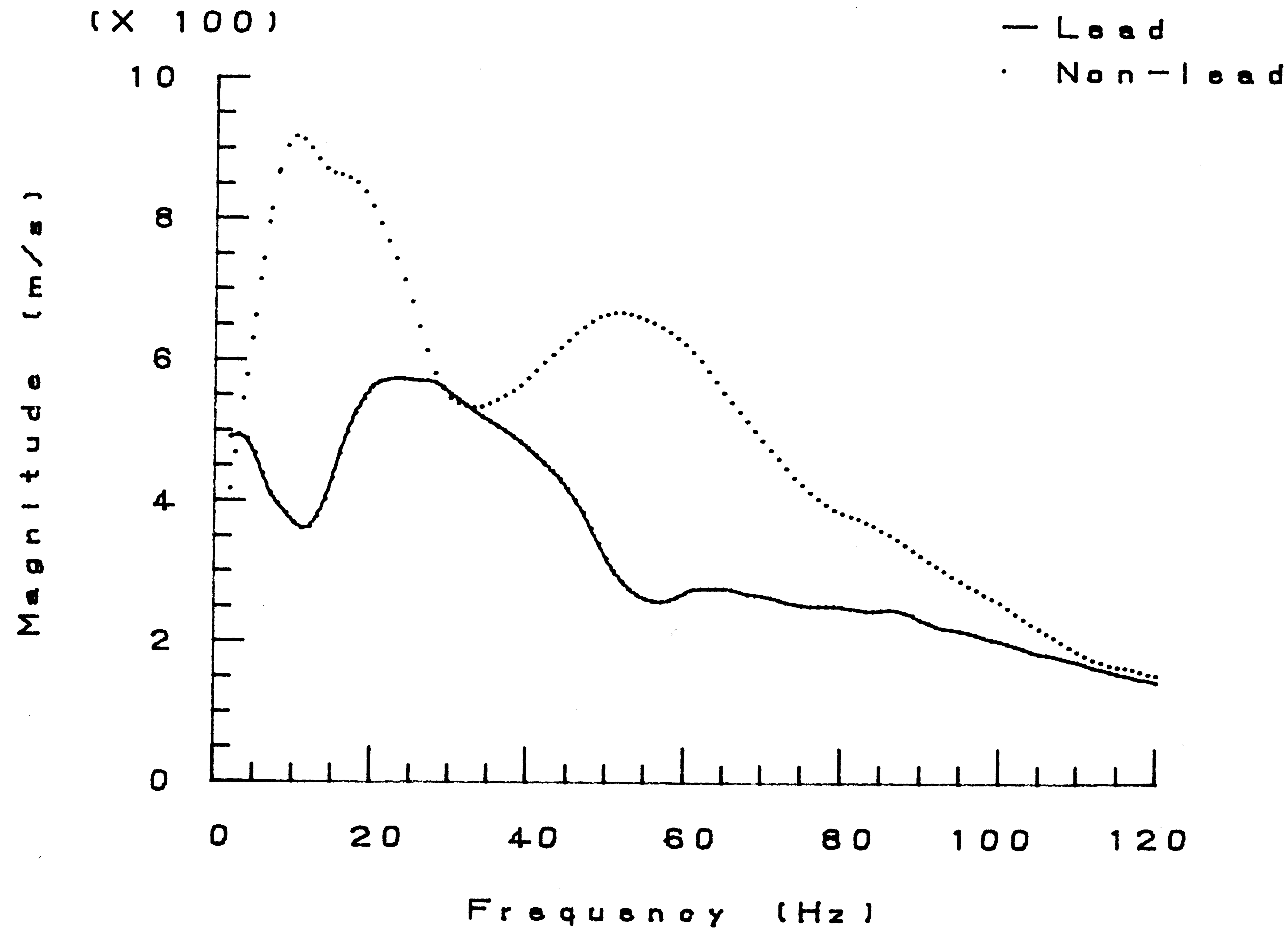

Fig. 19. Average spectral density of the lead and non-lead foreleg impacts at a gallop.

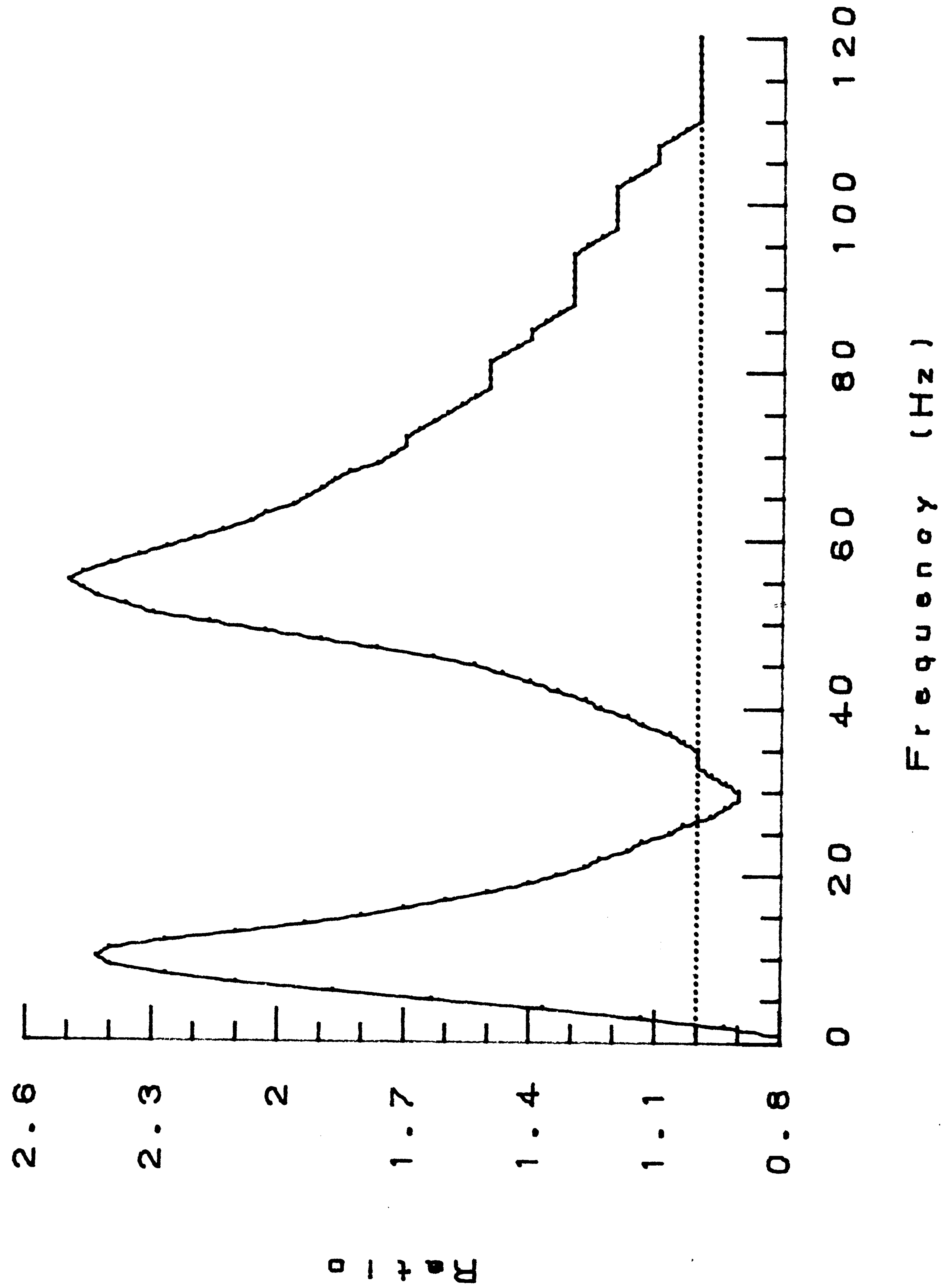

Fig. 20. Transfer function: Non-lead *vs.* lead foreleg impacts at a gallop.

12. Reduction of Impact Shock with Horseshoe Pads

It has been established that damage to the skeletal structure is closely dependent on the strain rate during loading. Behiri and Bonfield (1984), McElhaney and Beyers (1965), Lanyon and Smith (1970), and Lanyon (1971) all agree that strain rate has emerged as an important factor for the damage sustained by the bone and ultimately for the adaptation of the bone through the remodeling process.

Another failure model utilizes fatigue analysis to estimate the number of cycles or strides to failure. Pratt and O'Connor (1978) concluded that after 10 races, or $4 \bullet 10^3$ cycles, there would be a 40% drop in the cannon bone strength of a Thoroughbred. In a later study, Pratt (1982) looked at the effect of the different loading conditions as the racehorse moves from the straightaway to the turn. In the straightaway gallop, he estimated the cycles to failure of $\mathcal{N}=4310$ strides.

Pratt (1982) then used a shoe with pressure transducers to prove that while there was an even weight distribution on the stretches, almost all of the wieght was born by the outside of the shoe as it ran around the turn. Under these conditions, with an approximately 50% drop in the wieght bearing area of the joint surfaces, the number of cycles to failure dropped to $\mathcal{N}=42$, just $280m$.

Another fatigue analysis study was made by Nunamaker and Butterweck (1986). They collected *in vivo* data from strain guages installed on the third metacarpal of Thoroughbreds in training. They found an excellent fit of the strain data to a σ-$\mathcal{N}$ plot. They theorize that the constant activity of the bone cortex in remodelling allows for an operative definition of the endurance limit of the bone. In order to avoid catastrophic failure, there must be a balance between the rate of damage sustained from fatigue and the rate of bone remodelling.

This concept of a biological limit on the activities of an athlete makes intuitive sense and has been widely acknowledged for thousands of years although previously unproven. The use of a padding material between the horseshoe and the hoof supposedly reduces the amount of damage sustained and thus increases the amount of work that the animal can be safely asked to do. Whether the use of horseshoe pads actually contributes more to the peace of mind of the horse owner than to the skeletal structure of the horse is a matter of some debate. It is for these reasons that the shock absorbing capability of horseshoe pads was chosen as a topic for investigation.

Two different materials were chosen for study. Pad #1 was a viscoelastic material that had been proven to be an effective shock attenuating material when used as an insole for human shoes. The pad thickness was 3.17*mm*. Pad #2 was a commercially available horseshoe pad which had been used with good results on Thoroughbred racehorses by the author. This pad was more stiff than pad #1 and at 4.77*mm* was slightly thicker.

The basic testing routine was to start with no horseshoe pads and record both a left and a right lead gallop. The horse was then walked to the barn, his front shoes removed and then reset with the addition of pad #1. The horse was walked to the ring and again gallopped on each lead. This process was performed two more times, once with pad #2 and then again with no pad. The riding surface was a short distance from the barn and the entire procedure took less than four hours. The aluminum outer rim shoes were set and reset with size $3\frac{1}{2}$ race nails each time and no problems were encountered during or after the test with loose or shifting shoes or damage to the hoof wall.

Eighty peak impacts and 64 power density functions were averaged for each of the four trials. No significant difference was found between the first trial and the fourth suggesting that fatigue of the test horse had not had any effect. Each of the two pad materials proved to have a significant reduction of peak impact on one lead and no significant effect on the other lead. Pad #1 reduced the average peak impact acceleration of the non-lead foreleg by 7.2%. Pad #2 reduced the impact of the lead foreleg by 7.7%. A summary of the average peak impacts is given in Fig. 21. The transfer functions derived for the two cases with pads as compared to without pads are shown in Figs. 22 and 23.

Fig. 21. Summary of Results of the Use of Horseshoe Pads ($\mathcal{N}$=80).

Pad	Thickness (mm)	Lead Foreleg ave. ±std. dev.	%drop	Non-lead Foreleg ave. ±std. dev.	%drop
none	na	$11.9g \pm 3.6g$	na	$26.4g \pm 6.8g$	na
1	3.17	$12.1g \pm 3.5g$	0	$24.4g \pm 6.4g$	7.6%
2	4.77	$10.9g \pm 3.4g$	7.2%	$26.7g \pm 8.3g$	0

The size of these reductions may not seem to warrant the time and expense involved in the regular use of horseshoe pads but three important factors need to be considered. Firstly, all estimates of the safety factors based on the ultimate strength of the skeletal system of the racing Thoroughbred are small, usually close to 2. This means that even a 7% benefit from pads can have noticeably beneficial effects. Secondly, if the fatigue damage vs. bone remodeling theory of failure is used, reduction of the peak stress will have an exponential effect as determined by the logarithmic equation for cycles to failure.

Thirdly, a more rigorous experimental procedure would have used local digital nerve blocks to temporarily numb the horse's feet. It may be that a horse with horseshoe pads has a more aggressive stride while the same horse without the pads would extend less to spare himself some degree of impact shock. While the horse with the pads is getting the benefit of their protection, he would be hiding this fact from the researcher.

The fact that the two pads behave differently for the different leads is not suprising. In the last section, it was shown that each lead can be characterized by a different band of impact shock frequencies. The results in this section reflect the fact that shock absorbing materials often exhibit frequency dependent performance. It is not suggested and should not be inferred that the use of different horseshoe pads on different feet is recommended in this report.

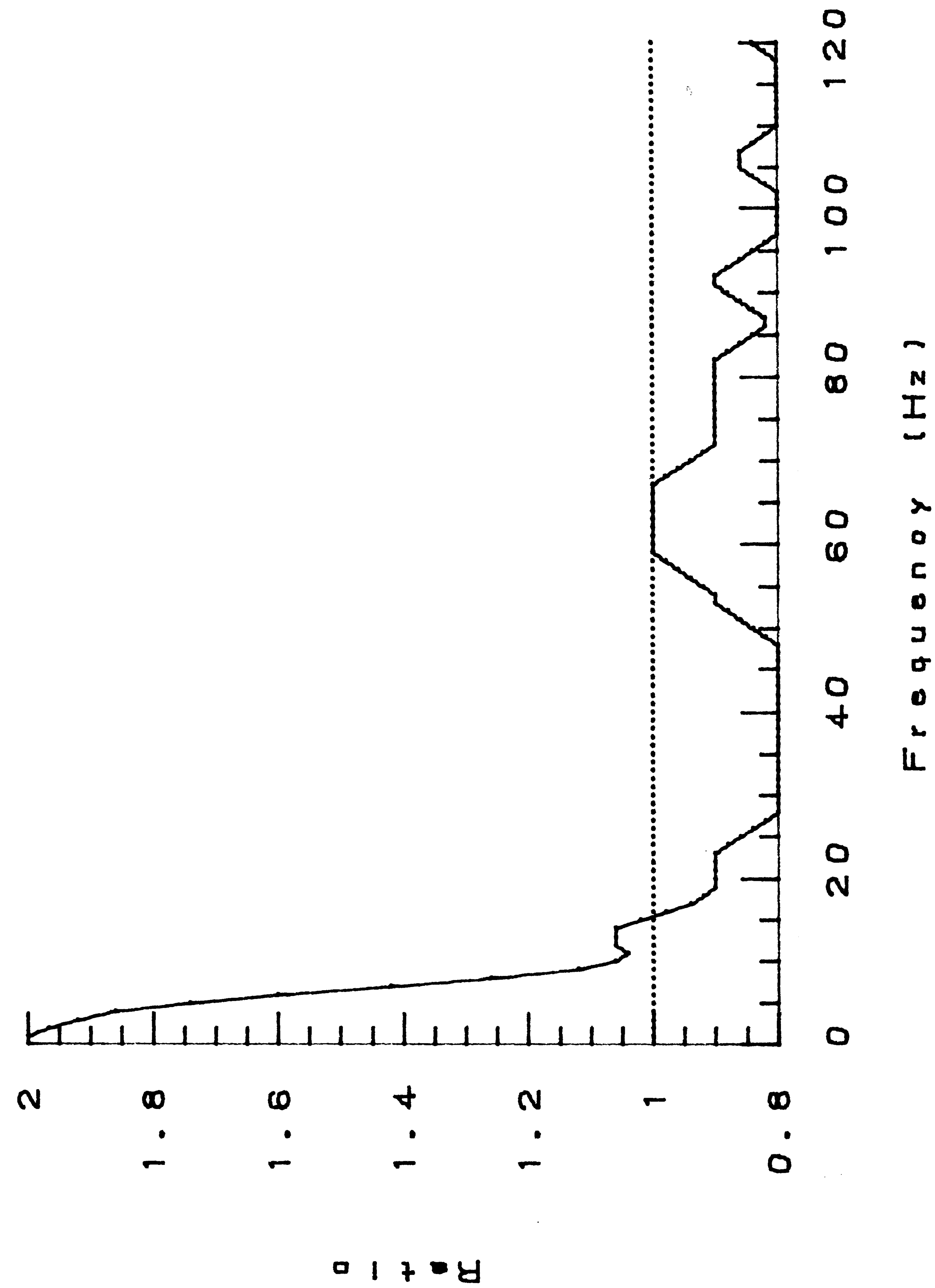

Fig. 22. Transfer function: Lead foreleg with pad *vs.* same leg without pad.

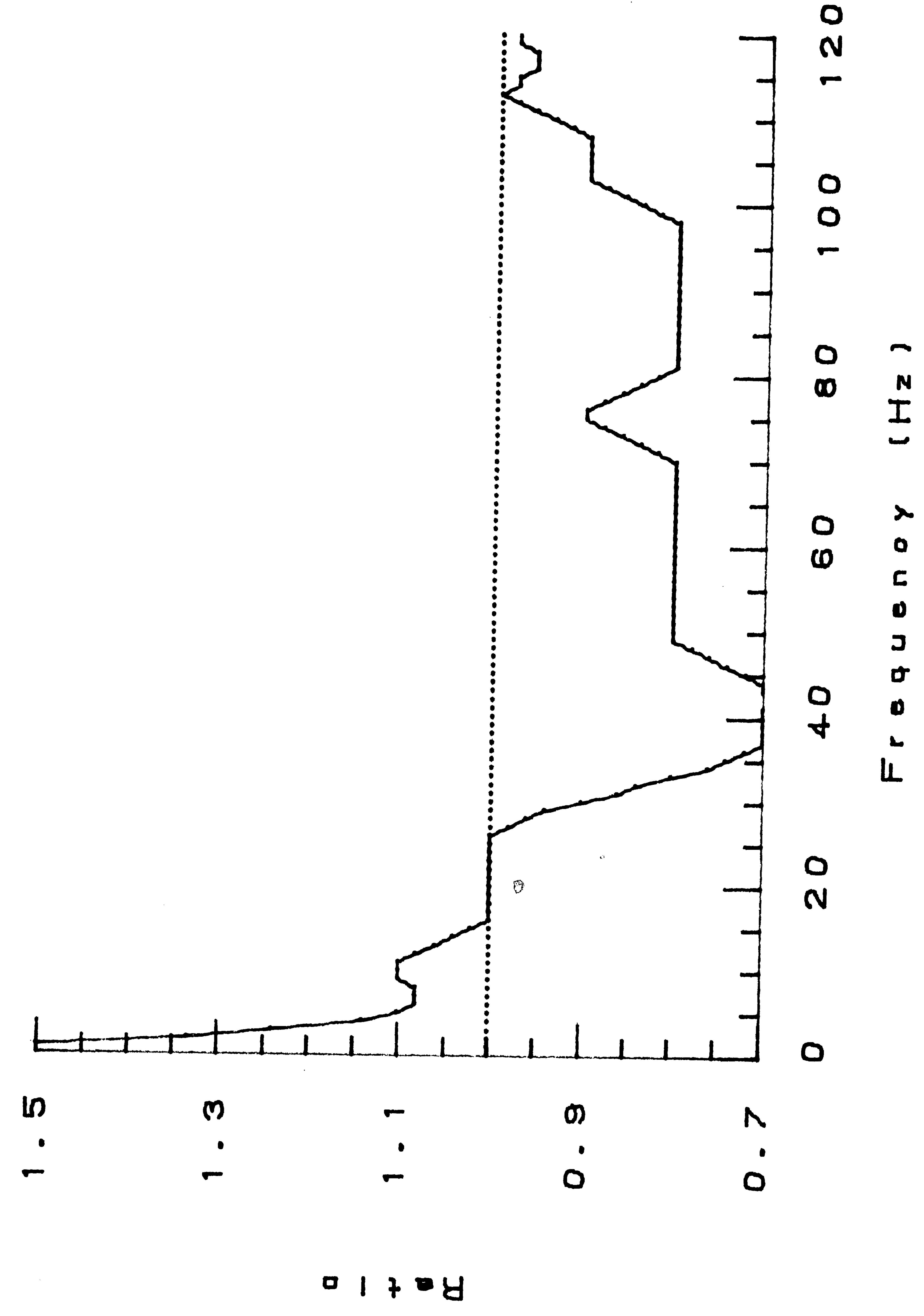

Fig. 23. Transfer function: Non-lead foreleg with pad *vs.* same leg without pad.

13. Influence of Racing Surfaces on Impact Shock

Studies by Cheney et al. (1973) and Arthur (1985) have shown that there is a connection between ground surface condition and lameness in the horses which train and race over the surface. Some material properties of the running surface are static or slowly varying, including surface content and depth, particle size and dry weight density. Other properties are constantly changing because of their dependence on weather patterns. Moisture content of the material will influence compactibility, density, and shear strength.

All of these factors will influence the dynamic response of the surface during impact, deceleration, support and propulsion. Cheney et al. (1973) used a mechanical testing devise to measure the response of California racetracks. They also surveyed horse trainers concerning injury rates and were able to successfully correlate ground reaction force with the rate of lameness. Frederick and Henderson (1970) showed graphically that changes in force-time patterns as a horse galloped over different surfaces.

Evidence also exists that the racing surface also has an influence on the type of injury. In general, as cushion depth increases, bone injuries will lessen and soft tissue injuries will increase but surfaces do influence injury in a more specific sense. Arthur (1985) spoke about a thorough remake of the work done by Cheney et al. (1973). As with the earlier work, that project was prompted by a remarkably high injury rate at local racetracks. In a four month racing period there were 43 cannon bone fractures, amounting to roughly 2% of all horses racing during that time.

Arthur (1985) went on to say that common occurrences of certain types of fractures definitely depends on the type of racing surface. While cannon bone fractures are the most common in the United States, they are relatively rare in England, where pastern fractures occur most frequently. This difference seems to be quite pronounced with the relative incidence being ten to fifteen both here and in England.

A series of experiments was used to test the applicability of the accelerometer system with regard to investigation of racing surfaces. The surfaces used were the dirt track and turf course at Penn National Race Course, Grantville, Pennsylvania. The same test horse was used as in the experiments described above and was shipped far enough ahead of the actual tests to allow for acclimatization and recuperation. All of the gallops were performed on an empty track after regular training hours. The cooperation of the management and maintenance personnel during these tests was invaluable.

The experiments were designed to look at differences between the dirt and turf courses and at the difference between the dirt track before and after harrowing. Just after the closing of the track, a single pass around the inside of the track was made with a harrow. The horse was then galloped $800m$ on the harrowed section, walked for $800m$ to the starting point, and galloped $800m$ on the unharrowed section. The two passes were made at approximately 3 and 5 meters from the rail. Soon after the trials on the dirt track, the horse was galloped on the turf course, again for roughly $800m$. Although the weather during the tests was seasonal for August, the weather for the week prior to testing had been extraordinarily hot and dry.

Peak impact accelerations were averaged for 110 impacts with the following results. The average peak on the harrowed section was $16.4g \pm 5.2g$, the average peak on the unharrowed section was $17.1g \pm 6g$, and the average peak on the turf was $17.8g \pm 5.3g$. For this discussion, a graphical representation of this data, the notched box and whisker plot (Fig. 24), will be used. In Fig. 24, the first box represents the dirt track before harrowing. The second represents the main track after harrowing and the third represents the turf course.

The central line in the box and whisker plot is the median value for that group of peak impact accelerations. The notch spans the 95% confidence interval of that median value. By comparing the notches in Fig. 24, it can easily be seen that each of the medians is within the confidence interval of the others. In other words, at the 95% confidence level there is no statistical difference between these medians.

The top and bottom of the box are located at the upper and lower quartile ranges of the data. The box thus covers the middle 50% of the data values. In normal cases, the whiskers extending from the box extend to the maximum and minimum values. In cases where there are extremely high or low values, the whiskers cover 1.5

times the interquartile range and the unusual values are plotted as single points.

In the 110 strides investigated in each case in Fig. 24, the harrowed dirt track and the turf track each had one such extreme value. The dirt track in unharrowed condition had five extreme impacts. It is well established that poorly maintained racing surfaces place the horse at increased risk of injury. These results suggest that some benefits of track grooming lie in the reduction of the number of harsh impacts through increased uniformity of the surface.

Another fact relating to surfaces and injury is that natural surfaces such as turf or wood chips in some way tend to produce less injuries in training and racing. A possible reason for this occurrence can be seen in the frequency domain comparison between the harrowed dirt track and the turf track. While there was very little difference in the power density of the dirt track before and after harrowing, there is an interesting band shift from higher to lower when the surface is changed from dirt to grass. The average power density of 64 strides for each case is shown in Fig. 25. The previous chapter discussed the dependence of bone damage on strain rate. In that respect, the shift from higher to lower frequencies of impact acceleration suggests a lower applied strain rate on the skeletal structure and a lower rate of catastrophic failure.

The derived transfer function for this comparison is shown in Fig. 26. Power density of the impact accelerations on the harrowed dirt track forms the basis and power density of impacts on the turf is used for comparison. Sixty impacts of the lead foreleg were averaged in each case. The horse obviously experiences a lower frequency shock wave when galloping on the turf.

A summary of the results of these tests can now be made. In the time domain, the average peak impact acceleration remained virtually unchanged. The dirt track before harrowing showed the highest percent standard deviation and five times as many extremely large peak impacts. The turf course produced lower frequency acclerations, in general, than the dirt. The use of this accelerometric system proved to be a simple means of quantifying the effects that changing conditions have on the horse.

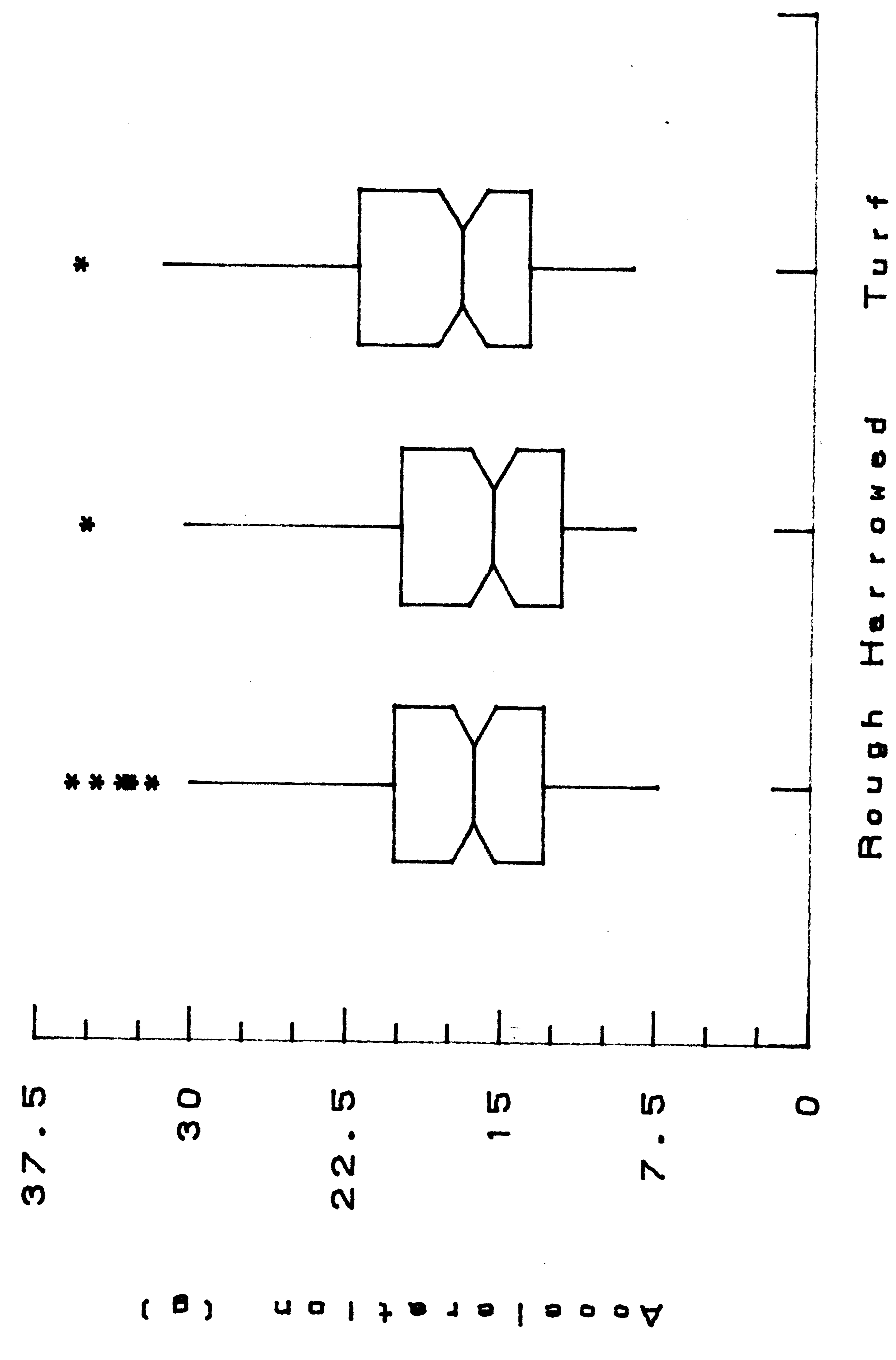

Fig. 24. Box and whisker plot: Main track before and after harrowing and turf track.

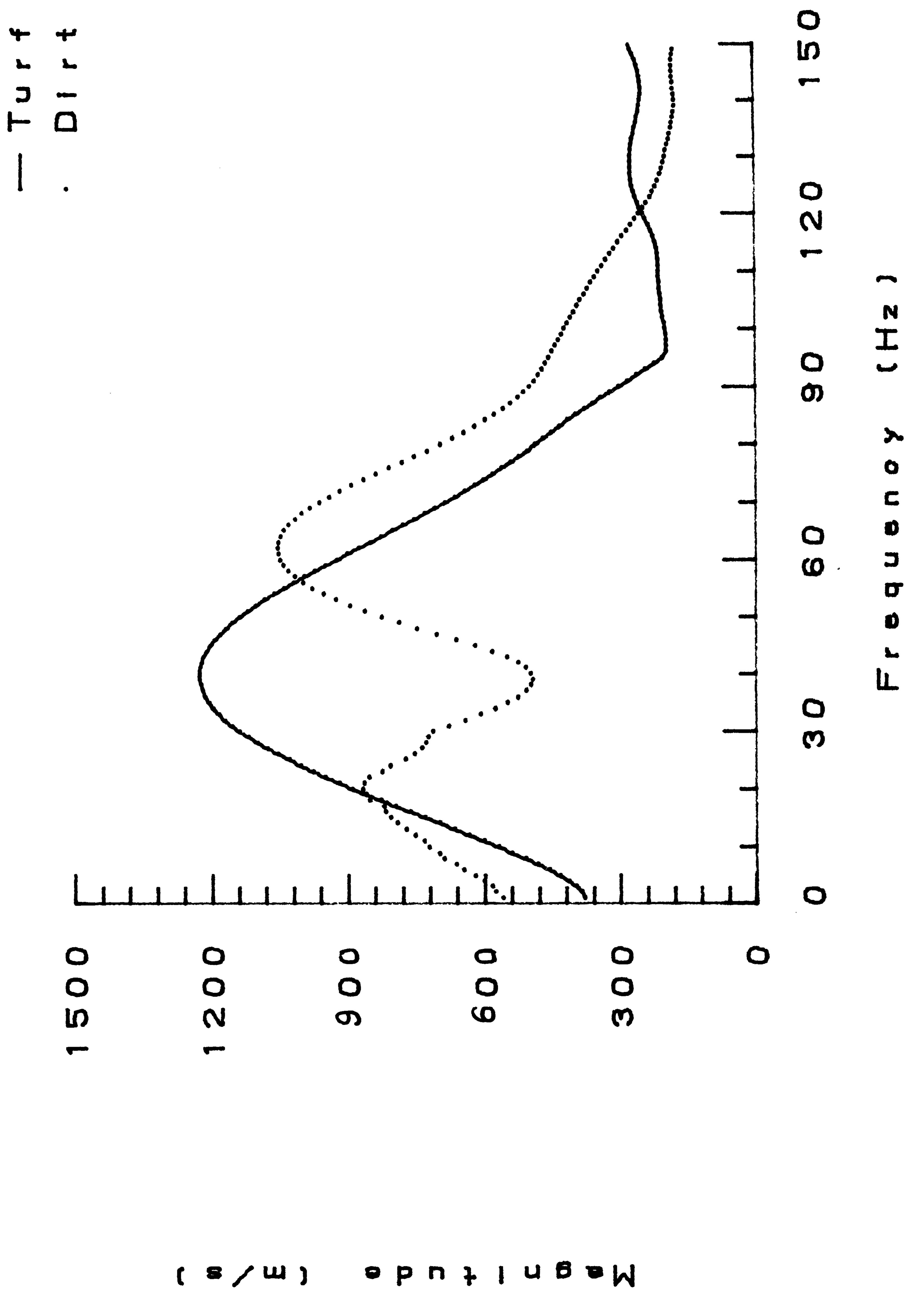

Fig. 25. Average spectral density of the lead foreleg impacts on the turf and on the harrowed dirt track.

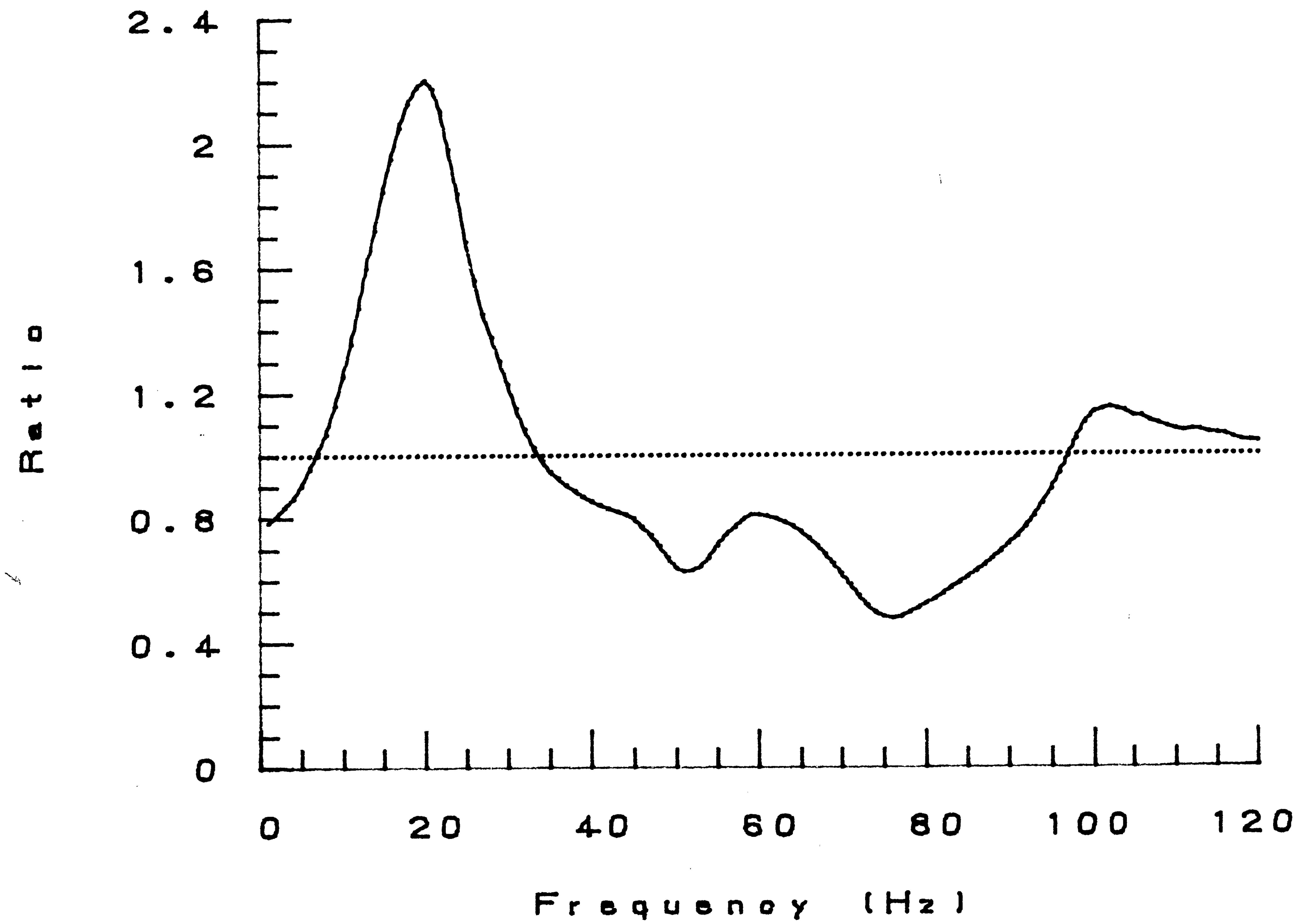

Fig. 26. Transfer function: Turf track *vs.* main track.

14. Measurements with Multiple Accelerometers

The analysis of intraskeletal shock attenuation and the calculations of internal forces of the lower leg are two examples of studies which require multi-channel data. The study of intra-skeletal shock attenuation involves measurements at several points and comparison of the accelerations in time and frequency domains. Any reduction of the energy of impact shock between those points would then be due to dissipation, energy conversion, or energy storage. The experiments of Loy (1987) show that these processes exhibit frequency dependence.

The calculation of internal forces in the lower leg can be useful in a wide range of studies. These studies typically involve finding of forces, strains, or strain rates of a particular structure. Kingsbury et al. (1978) used dead legs and mechanical loading devices which reproduce the ground reaction forces of a moving horse. A more elegant method of finding the internal forces is to model the lower leg as done by Bartel et al. (1978). Work in the area of modelling the human locomotion system is in progess at Lehigh University and necessarily involves the use of multi-channel accelerometric data to provide the time dependent boundary conditions required for solution and verification of the modelling equations.

A test was performed using three accelerometers. The first was attached to the hoof as described previously. The second was fixed externally to the midpoint of the third metacarpal bone of the horse. The third was fixed externally to the forehead of the rider. The methods for external mounting were developed by Loy (1987) and utilize elastic wrappings to provide simple non-invasive attachment of an accelerometer. The purpose of the present experiment was to develop and evaluate procedures for collecting multi-channel *in vivo* data under stable conditions.

The time required to install the accelerometers and secure the cables increased slightly for this test. Keeping the accelerometer cables safe and clean until they are secured and the slack taken up requires an extra helper but an efficient routine had been estblished during the single-channel testing. The same ground surface was used as for the testing discussed in chapters 9,11, and 12. The average peak accelerations as measured at each point is shown in Fig. 27.

Fig. 27. Summary of results with multiple accelerometers ($\mathcal{N}$=50).

Gait	Peak Acceleration hoof	leg	rider
Foreleg at a walk	$7.2g \pm 3.0g$	$7.0g \pm 2.6g$	na
Foreleg at a trot	$9.8g \pm 3.3g$	$9.2g \pm 2.3g$	$2.8g \pm .8g$
Lead foreleg at a gallop	$14.8g \pm 4.3g$	$14.1g \pm 2.9g$	$3.5g \pm 1.1g$
P value	$< .0001$	$< .0001$	$< .0002$

Comparing the measurements, several points can be made. First, the average peak impact accelerations do not significantly differ between hoof and leg at any of the three gaits. This indicates that the shock wave travels through the lower leg with virtually no attenuation. The differences in peak acceleration do have a highly significant variation ($P<.0002$) when comparing the average peak impact between gaits at the same measurement point. Also, the differences between the measurements from the rider differ significantly ($P<.0001$) from the measurements from both hoof and leg. Next, the standard deviation of the mean from the leg readings are over 20% lower than the standard deviation from the hoof readings. Apparently, the horse's mechanical system has some way of dealing with widely varying impacts.

This is shown graphically by the box and whisker plot in Fig. 28 where the median values are almost identical but the 95% confidence interval for the mean, the interquatile range covered by the box, and the range covered by the whiskers are all much more narrow for the peak impacts measured from the leg. Fig. 29 shows two separate gait patterns as measured at the hoof and leg. As with the previous plots of time domain signals, the first large peak is impact and the second large peak is take-off.

Data from the rider's head suffered from poor resolution at the walk but shows good results at the trot and gallop. At the faster gaits, there was a significant attenuation of the impact shock wave between the lower leg and the rider's forehead. When comparing the data from the rider's forehead to that of the hoof or leg, it is seen that the combination of horse, tack, and rider contributed to a total of 70-75% reduction in the average peak impact acceleration.

The accelerometric system works well for multi-channel measurements. It is only slightly harder to complete all connections to power packs, accelerometers, and to the recorder. The additional weight carried by the rider was of little concern. The results from this use of the system are encouraging since the modelling of the equine locomotion system and the study of intraskeletal shock absorption are two areas with great potential.

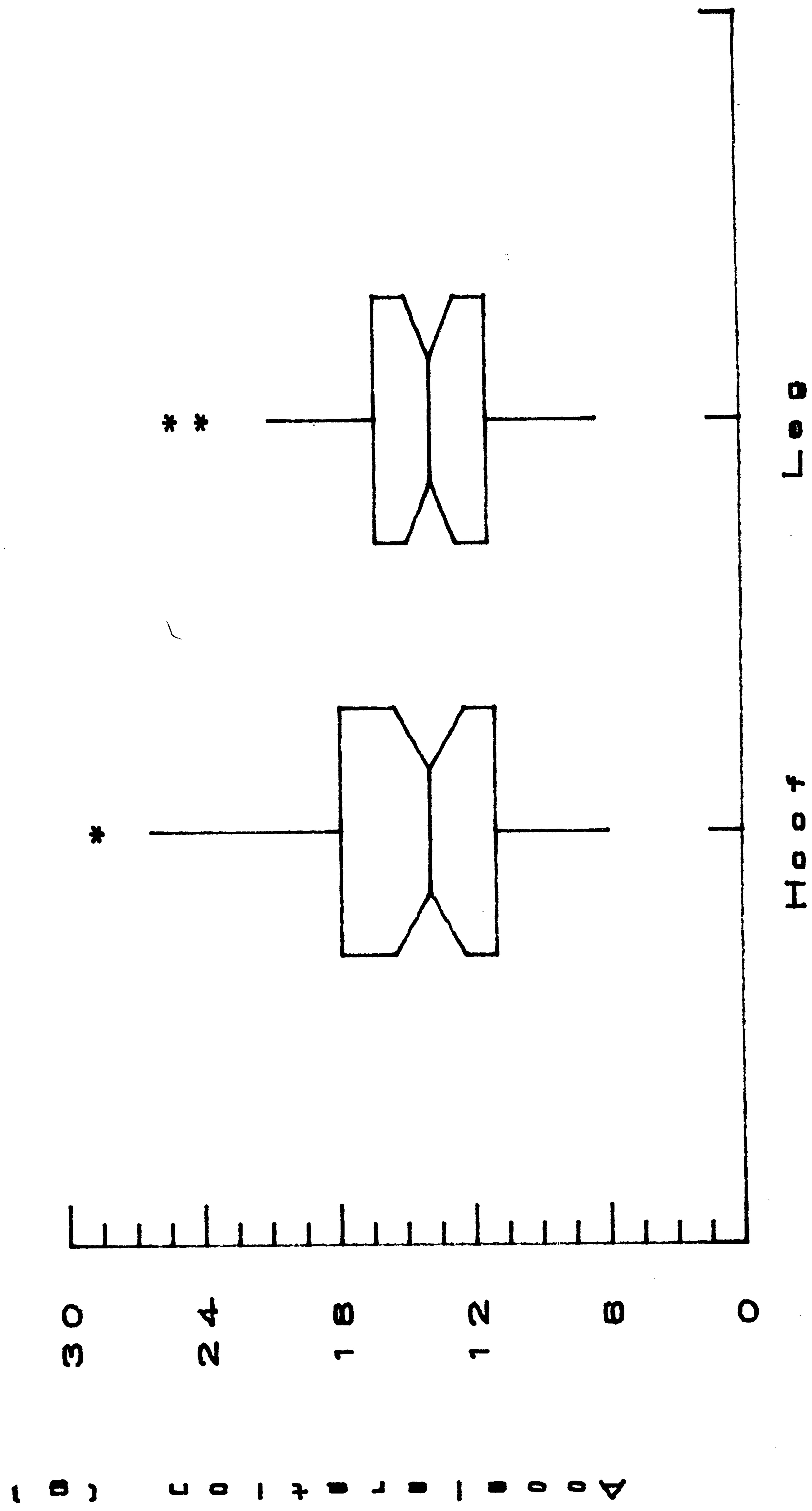

Fig. 28. Box and whisker plot: Impacts measured from hoof and leg.

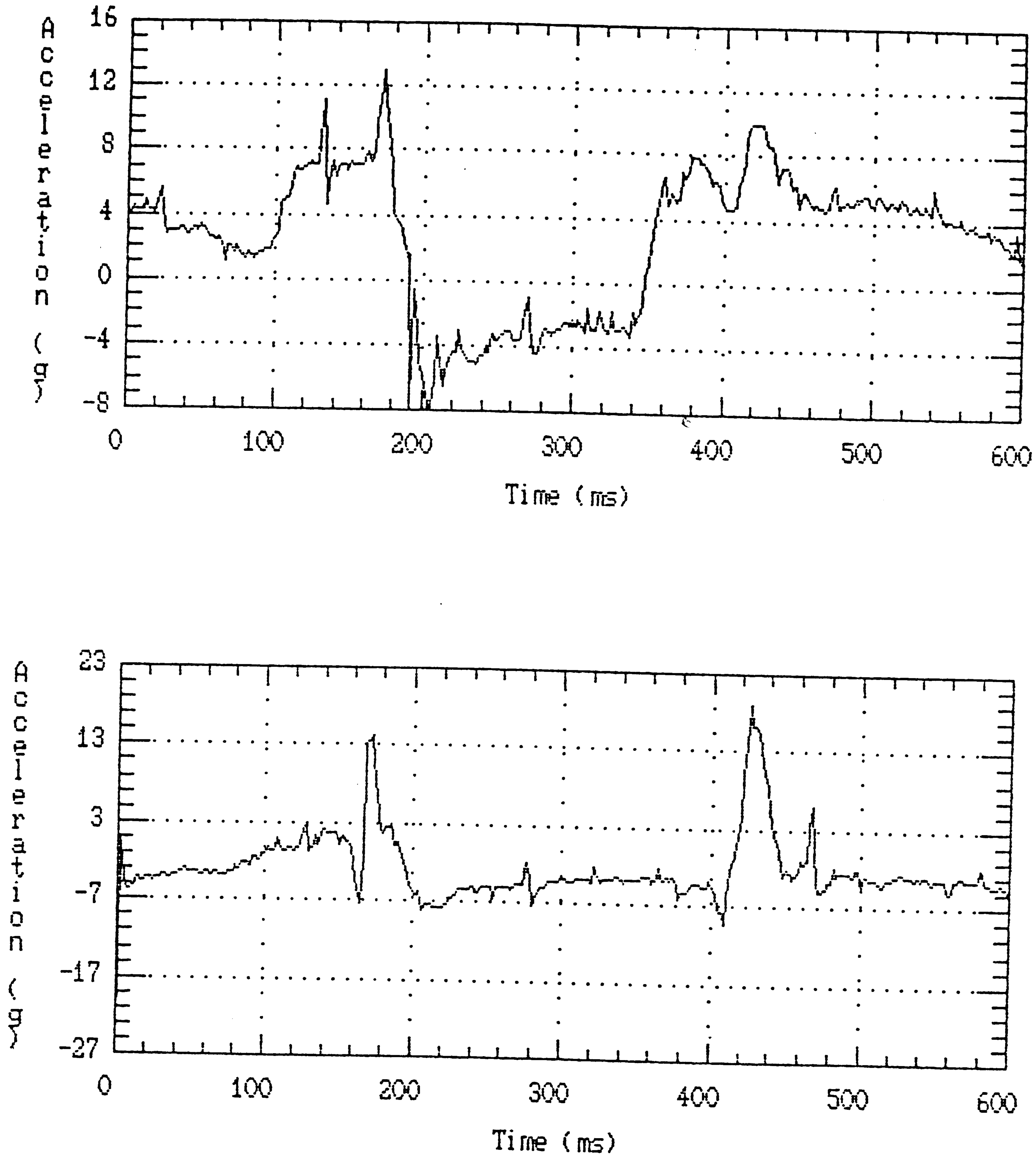

Fig. 29. Temporal gait pattern: Impact measured at lead foreleg third metacarpal (top) and hoof (bottom) at a gallop.

15. Conclusion

The locomotion of horses is a very complex pattern of flight, impact, suspension, and take-off. No single method can totally describe this behavior and several methods have been discussed along with their advantages and disadvantages. Force plates methods are most common and promising results have been obtained in various research areas, particularly with regards to lameness diagnosis. For this study it was decided to use *in vivo* testing with on-board recording because of the experimental advantages and the wide range of potential study.

Recognising that the impact phase plays an important role in the application of forces to the biomechanical structure of the horse, it was the purpose of this research to develop and evaluate methods for the investigation of the conditions which occur at impact. The system described in this report proved to live up to all expectations and has great potential for further development.

Accuracy of the system is one area which could use further work, as there were some errors introduced by the recording and digital conversion of the accelerometer signals. Errors in data recording could be minimized with routine cleaning and demagnetizing of the recorder. The digital conversion harware had an adjustable range which allowed the resolution to be maximized. Except for the experiment discussed in chapter 14, resolution was kept at an optimum. The estimated experimental error for these results is less than 3% at the gallop. In chapter 14, experimental error from recording and conversion is estimated to be on the order of 10%.

These errors were not of great concern since they were consistent in nature and could be minimized and accounted for. Since they are related to the experimental hardware and not to experimental design, they could conceivably be eliminated with advances in instrumentation. Also important was the fact that the inherent variability from live subjects under field conditions prevents rigorous data treatment. That is that since some factors change from day to day, we are limited to making relative comparisons between large statistical pools and thus don't require more accurate measurement capability.

The reliability of the system was proven by the consistency and reproducibility of the results. The tests described above represent data collected over twelve months with excellent repeatability throughout. This is shown in Fig. 30, where average peak impact accelerations from the lead foreleg at a gallop are compared over the twelve month period.

Time and frequency domain patterns of the natural gaits were reproduced over and over again to the point where they became recognizable and were used to characterise the gaits. With the exception of two cable failures, the system always produced consistent results under stable conditions. This is very important because the data was recorded without on-line monitoring to check for errors. Without a highly reliable system, much time would be lost as entire experiments are done over.

Fig. 30. Average lead foreleg impact at the gallop in tests over one year's time.

Page	Number of trials	Average	Date	Condition
61	$\mathcal{N}=80$	$11.9g\pm3.6g$	6/10/87	dirt riding ring
68	$\mathcal{N}=110$	$16.4g\pm5.2g$	7/23/87	harrowed dirt track
80	$\mathcal{N}=50$	$14.8g\pm4.3g$	5/07/88	dirt riding ring

In each of the experiments discussed, the system of data collection and analysis had sufficient resolution to provide useful information concerning the conditions of the test. The experiments in this study were chosen to represent as wide a range as possible for applications of these techniques. The success of the accelerometric system is therefore encouraging and suggests many possible areas for in-depth study of impact conditions relating to equine locomotion.

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Vita

Michael J. Platt, the son of Benjamin and Ellen Platt, was born on September 30, 1960 in Philadelphia, Pennsylvania. In 1982, he was graduated from Colgate University, Hamilton, New York with a B.A. in Physics. Later in the same year, he recieved training as a horseshoer at the New York State Veterinary College, Cornell University, Ithaca, New York. He worked as a professional farrier in Southeast Pennsylvania, served a racetrack apprenticeship, and eventually specialized in Thoroughbred horseshoeing. Married to Emily West Platt, he currently resides in Auburn, Pennsylvania.